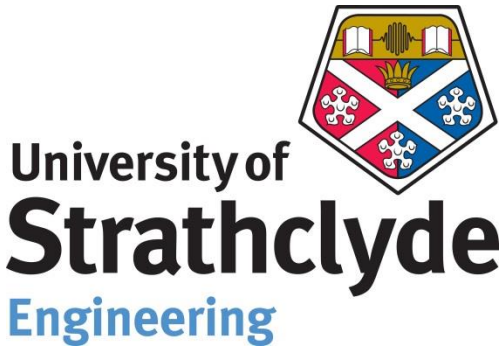

Department of Biomedical Engineering

Visual Feedback in Orthopaedic Rehabilitation

By

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This thesis is submitted in fulfilment of the requirements for the degree of PhD in Biomedical Engineering

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Copyright Statement

Signed:

Date:

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Abstract

Currently, functional outcome following total knee arthroplasty (TKA) surgery is often not restored, with the majority of TKA patients exhibiting lower functional outcome scores than healthy counterparts. There is some controversy regarding the nature of rehabilitation delivery following TKA surgery which could contribute to sub-optimal outcomes. Visual feedback has had a positive effect in other patient populations, such as stroke survivors, and therefore may also improve the efficacy of TKA rehabilitation. Currently, the most effective way to deliver visual feedback is with motion analysis technology. However, current protocols are not suitable for routine clinical use as they are time consuming and complex. Therefore, the aims of this study were to develop a motion analysis protocol tailored for routine clinical use, use the protocol to implement real-time visual feedback to TKA patients and test the effectiveness of the feedback on patients' functional outcome.

A cluster based protocol was developed (Strathclyde Cluster Model; SCM) and compared to the current clinical gold standard (Vicon Plug in Gait; PiG) in terms of kinematic output and inter/intra-assessor reliability. SCM was used to implement 3 visual feedback scenarios during TKA rehabilitation. To test the effectiveness of visual feedback, functional outcome was compared for a group of patients who received feedback and a group of controls. Further, the acceptability and reliability of SCM was tested with clinicians who had no prior experience in motion analysis.

Results demonstrated that SCM was generally as reliable and accurate as PiG. Further, visual feedback does appear to have a positive effect on TKA patients and when tested with clinicians who were inexperienced in motion analysis, SCM was generally acceptable and reliable.

In conclusion, SCM is an appropriate protocol for routine clinical use to deliver visual feedback during TKA rehabilitation and visual feedback has a positive effect on outcome for TKA patients.

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List of Abbreviations

ACL	Anterior Cruciate Ligament
AJC	Ankle Joint Centre
ARF	Anatomical Reference Frame
ASIS	Anterior Superior Iliac Spine
CA	Calcaneus
CMC	Correlation of Multiple Coefficients
CoM	Centre of Mass
DoF	Degrees of Freedom
FHS	Flexion at Heel Strike
FM	First Metatarsal
FMS	Flexion at Midstance
GA	Gait Analysis
GRF	Global Reference Frame
GRV	Ground Reaction Vector
GS	Gait Symmetry
HBM	Human Body Model
HH	Helen Hayes
HIR	Hip Internal Rotation
HJC	Hip Joint Centre
IC	Initial Contact
ICC	Intra Class Correlation
KAM	Knee Adduction Moment
KJC	Knee Joint Centre
LE	Lateral Epicondyle
LLVT	Lower Limb Visualisation Tool
LM	Lateral Malleolus
LR	Loading Response
ME	Medial Epicondyle
MM	Medial Malleolus
OA	Osteoarthritis
OKS	Oxford Knee Score
PEVS	Peak Extension Velocity in Swing
PFLR	Peak Flexion Loading Response
PFS	Peak Flexion in Swing
PiG	Plug in Gait
PRS	Peak Rotation in Swing
PSIS	Posterior Superior Iliac Spine
QoL	Quality of Life
ROM	Range of Motion
SCM	Strathclyde Cluster Model
SD	Standard Deviation
SI	Symmetry Index
SPT	Spatiotemporal
STA	Soft Tissue Artefact
STS	Sit to Stand
TKA	Total Knee Arthroplasty

TRF	Technical Reference Frame
VM	Fifth Metatarsal
VR	Virtual Reality

Chapter 1: Introduction and Literature Review

Chapter 1

1 Introduction and Literature Review

Rehabilitation is defined as 'restoration to health or normal life by training and therapy' (Oxford English Dictionary). In the clinical environment, rehabilitation is a therapeutic programme designed to restore function following impairment. Emphasis is placed on:

- Preventing contractures
- Developing muscle strength
- Training the patient to use any residual function in an effective way
- Guiding the patient and family in what is likely to be an altered way of life

Rehabilitation is not necessarily curative, but aims for effective performance for re-entry into society at a level which suits the needs of the patient (Nickel and Botte, 1992). Historically, rehabilitation was often viewed as the 'third phase of medicine', after prevention and treatment. However, it was soon recognised that a number of complications were caused by immobility following surgery and therefore remobilisation quickly became part of acute management of a number of conditions (Gordon, 1993). The general rule for when a rehabilitation programme should begin is as soon as possible after the impairment has occurred and as soon as the patient is physically able (Nickel and Botte, 1992).

There are a number of ways rehabilitation can be delivered, but the ultimate goal is to restore maximal function to the patient. There is some controversy surrounding how this is best achieved. Currently, depending on the capabilities of the patient, a

large amount of rehabilitation is carried out in the home setting, with minimal or no physiotherapy supervision, which is currently known as self-management (Kramer et al., 2003; Rajan et al., 2004). Some centres offer inpatient and/or outpatient rehabilitation programmes which are generally delivered by physiotherapists and focus on repetitive functional exercises assessed using observational methods or manual goniometers (Lingard et al., 2000; Toro et al., 2003). It has been suggested that this may not be the most effective way to measure performance and functional improvement, as observational assessment of movement can be subject to high inter and intra-assessor variability (Kawamura et al., 2007; Ong et al., 2008). Since its commercialisation, motion capture technology has been the gold standard for measuring human movement. Historically, motion analysis using this type of technology was restricted to research environments and complex clinical cases. However, the introduction of computer graphic imagery in the animation and gaming industries increased the demand for cheaper and more widely accessible motion capture technology. As this technology becomes more widely available, the possibility for its use to become part of routine clinical practice is ever increasing.

However, there are still a number of barriers preventing routine clinical use of motion analysis. Data collection and analysis with current commercially available systems is time consuming and complex (Toro et al., 2003). Additionally, feedback of results to patients and clinicians is not delivered in real-time. Post-processing is required to distil useful information from data collected which requires technical expertise from the user. Figure 1.1 demonstrates a typical workflow for an instrumented motion analysis session and highlights the technical proficiency that may be required to run the session effectively.

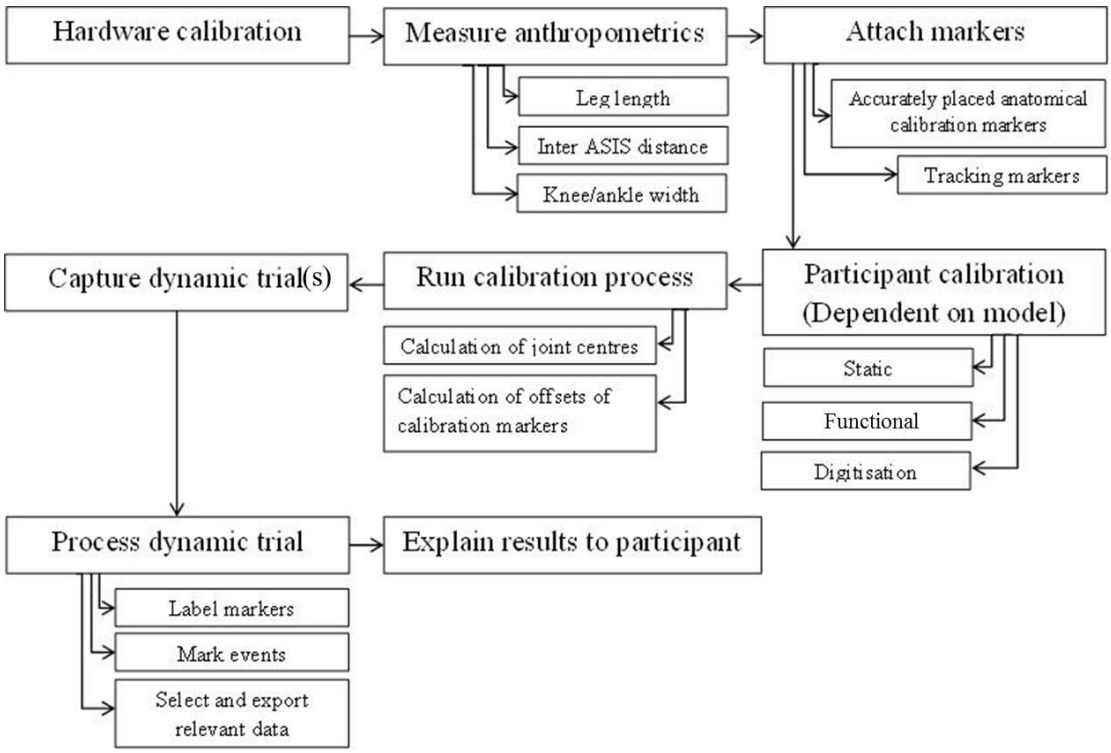


Figure 1.1 Typical motion analysis session workflow. ASIS – anterior superior iliac spine

One aspect of the motion analysis workflow which can be particularly difficult is the feedback of results to patients. It has been documented that patients don't respond well to figures and graphs generated by commercially available systems as many have difficulty understanding the information in this format (Loudon et al., 2012). Feedback of results to the patient is crucial in order to gain maximum effectiveness from the session and achieve maximal functional outcomes. Further, for learning or re-learning patients need timely knowledge of performance and results (Loudon et al., 2012). Virtual reality can provide patients with feedback on the quality of their movement, as well as a stimulating environment. Additionally, the use of virtual reality allows for so called 'purposeful gaming', which can increase patient compliance and satisfaction (Laver et al., 2011). While a number of authors (Chao et al., 2015; MacDonald et al., 2009; Holden, 2002; Webster and Celik, 2014; Wingham et al., 2015) have investigated the effectiveness of virtual reality and

purposeful gaming, the use of these systems in the clinic is still limited by technical inaccessibility of the motion capture technology and lack of clinical validation of virtual reality environments for rehabilitation.

As mentioned previously, there are currently inconsistencies regarding rehabilitation delivery which can lead to reduced functional outcomes for a number of patient populations. One area of rehabilitation which continues to deliver poorer functional outcomes than healthy controls is rehabilitation following total knee arthroplasty (TKA; Ouellet and Moffet, 2002).

1.1 Structure of the thesis

Due to the different nature of this work, the literature review has been divided between chapters. This was to allow the reader to more clearly appreciate the rationale behind chosen methodologies for each section.

Initially, TKA and TKA rehabilitation is discussed in a short literature review. Subsequent literature relating to each chapter is then presented in the introduction section of each chapter. Chapters 3 to 5 discuss the development of a biomechanical model for measuring kinematics and functional movement and therefore literature regarding gait analysis and biomechanical models is presented. Chapter 6 presents work on development of bespoke visualisations and therefore the literature regarding visual feedback is presented here. Chapter 7 presents the effectiveness of a visual feedback intervention on patients who have undergone TKA and therefore literature relating to this topic is presented here. Chapter 8 discusses reliability of using a motion analysis protocol in a clinical environment and therefore literature is provided on current clinical movement analysis methods and the repeatability and reliability of these methods.

1.2 Total Knee Arthroplasty

TKA is often the only treatment option for late stage osteoarthritis (OA). OA of the knee is the most common joint disorder in the UK (Duffell et al., 2014; Simon, 1999) and the prevalence of OA increases with increasing age, with 44% of people aged 80 or over displaying symptoms compared to 27% of people under the age of 70. There is also a higher prevalence in women compared to men, and obese people are nearly 3 times more at risk than people of normal weight (Felson et al., 1987; Suri et al., 2012). Radiographs are commonly used to determine the presence of OA and the Kellgren-Lawrence (K-L) scale is often used to determine the level of cartilage degeneration (Suri et al., 2012). According to the K-L scale, OA is determined from the following radiological features: formation of osteophytes; narrowing of joint cartilage associated with sclerosis of subchondral bone and altered shape of bone ends. The scale ranges from 0 (none) to 4 (severe) (Figure 1.2; Kellgren and Lawrence, 1957).

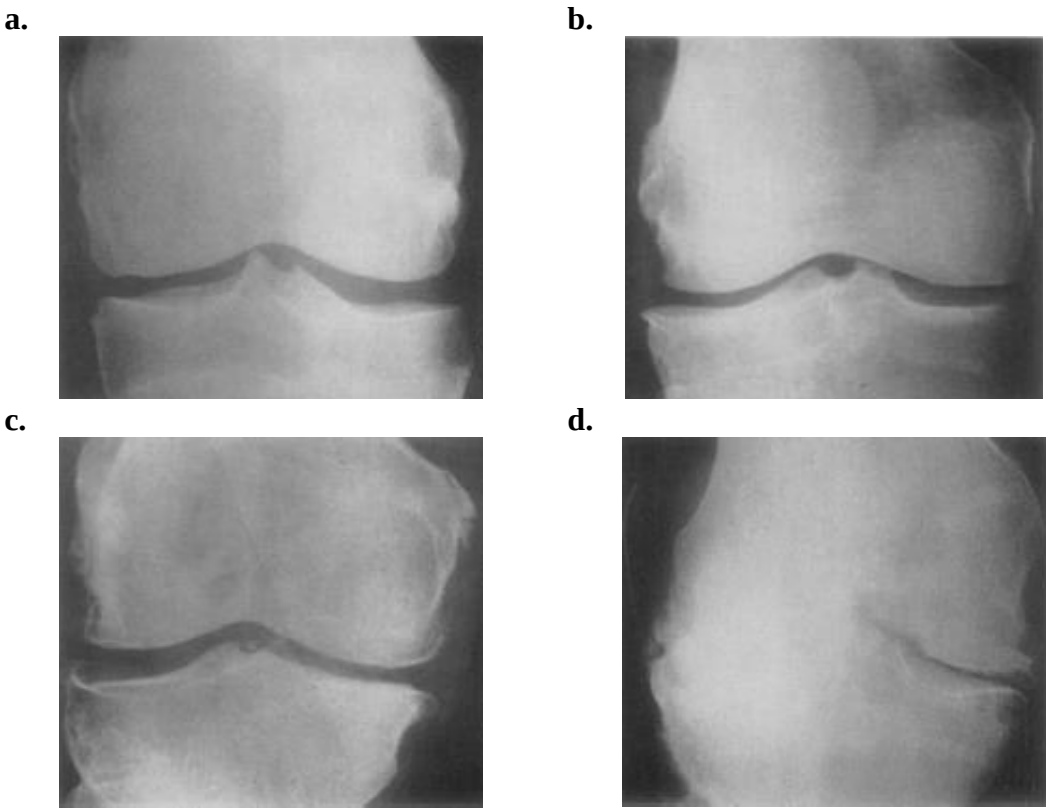


Figure 1.2 Radiograph of stages of knee OA as described by the Kellgren and Lawrence scale **a.** Grade I **b.** Grade II **c.** Grade III **d.** Grade IV (Kellgren and Lawrence, 1957)

While conservative treatment may reduce pain and improve mobility initially, joint replacement remains the only option for late stage knee OA (Liddle et al., 2013). The following section outlines the structure and function of the normal knee and disease progression of OA in the knee.

There are a number of different joint classifications depending on the anatomical structure of the joint. Fibrous joints are held together by ligaments whereas cartilaginous joints are joined by the cartilage which lies between them, for example, the vertebrae of the spine. The third and most common joint type is the synovial joint. Synovial joints make up the majority of joints in the body and are enclosed in a capsule which contains synovial fluid. Joint movement facilitates flow of synovial fluid which ensures the maintenance of healthy cartilage. The knee will be discussed in more detail as an example of a synovial joint (Figure 1.3). The femur, tibia and

patella comprise the knee joint. However, the femur and the tibia are the only bones which participate in articulation; the patella is a sesamoid bone which contributes only to stability of the knee (Senavongse et al., 2003). The articular surfaces of the femur and tibia are covered by hyaline cartilage which facilitates smooth articulation during movement. Also present within the joint capsule are the menisci which are fibrocartilagenous discs located on the proximal surface of the tibia. The menisci act to dissipate load through the knee during compression and also contribute to joint stability (Walker et al., 2015).

a. **b.**

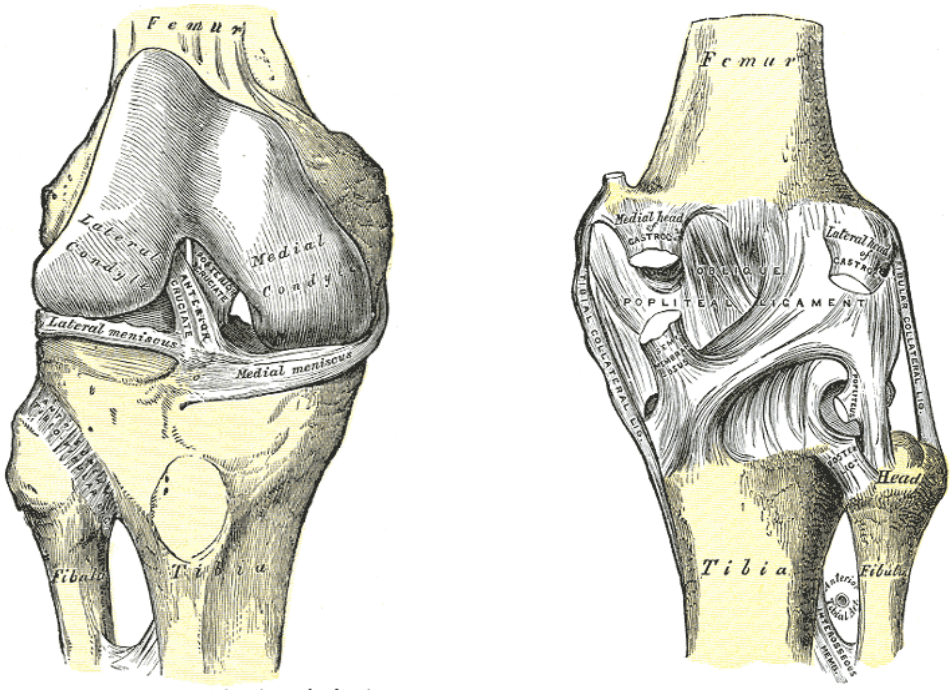


Figure 1.3 Anatomy of the knee showing the ligaments, menisci, femur, tibia and fibula. **a.** Anterior view **b.** Posterior view (Gray, 1918)

There are a number of ligaments involved in stability of the knee. The anterior and posterior cruciate ligaments (ACL and PCL) are located within the joint capsule and contribute to the anterior-posterior and rotational stability of the knee. Out with the joint capsule are the medial and lateral collateral ligaments (MCL and LCL) which

contribute to the medio-lateral stability. The MCL is stronger than the LCL as the knee is more likely to experience high forces in the medial direction.

Inadequate circulation of synovial fluid within the joint capsule or trauma can cause breakdown or damage of cartilage. Cartilage is non-regenerative and aneural, meaning that a sufferer of OA may be asymptomatic until the cartilage breakdown has reached the level of the underlying bone. By this point, the cartilage is so worn down that restoration to a healthy level is highly unlikely. The breakdown of cartilage can eventually lead to adjacent bones coming into contact with each other during movement which causes severe pain and formation of osteophytes (Figure 1.4).

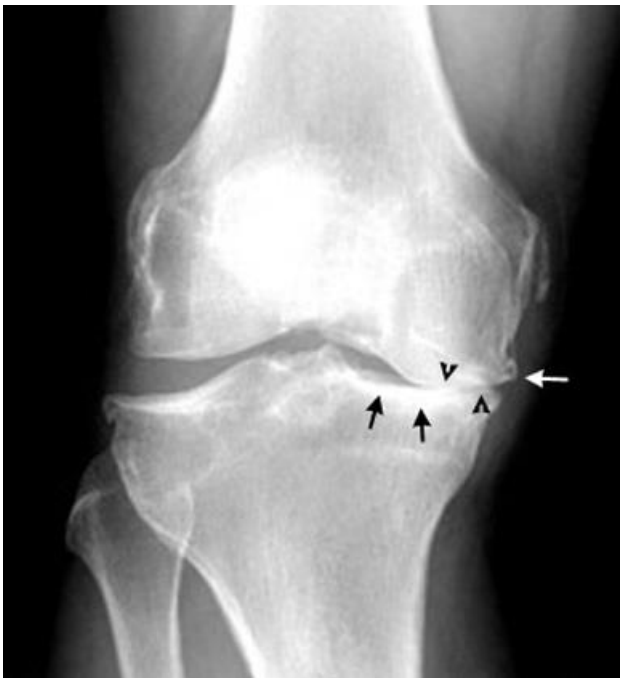


Figure 1.4 Radiograph of a knee with OA. The white arrow indicates formation of osteophytes, black arrows indicate cartilage degeneration and the black arrow heads indicate joint space narrowing

Knee arthroplasty is one of the most common operations in orthopaedic surgery worldwide (Drexler et al., 2013) and is often indicated in OA cases. The major aims of this type of reconstructive surgery are to relieve pain and restore stability and

functional movement to the joint (Laskin, 2008). Depending on the extent of the damage, a patient may be recommended for either Unicompartmental Knee Arthroplasty (UKA) or Total Knee Arthroplasty (TKA). UKA requires replacement of one of the condyles of the knee, whereas TKA requires replacement of both condyles.

UKA was previously seen as a suitable alternative to TKA in cases where complete replacement of the joint was not indicated. However, it has been suggested that TKA is now a more predictably successful operation with more reliable results than UKA (Bonin and Chambat, 2008). Despite the high success of TKA procedures, there are still a number of advantages of UKA over TKA. Patients tend to recover faster from UKA surgery, they tend to suffer from fewer complications and there tends to be lower morbidity resulting from surgery. Despite these advantages, there are also a number of complications as UKA implants exhibit a higher rate of early failure than TKA implants. To be eligible for UKA surgery, the patient must have limited joint degradation and all surrounding ligaments must be intact. Further, correct positioning of the implant can be difficult and is largely based on the skill of the operating surgeon. As such, indications for UKA are limited and its success is largely dependent on careful patient selection and implant design (Bonin and Chambat, 2008).

As a result of this, there is high demand for TKA surgeries and implant design has been constantly evolving since the 1950s (Bonin and Chambat, 2008). The majority of modern implants are composed of a femoral component and a tibial tray (Figure 1.5).

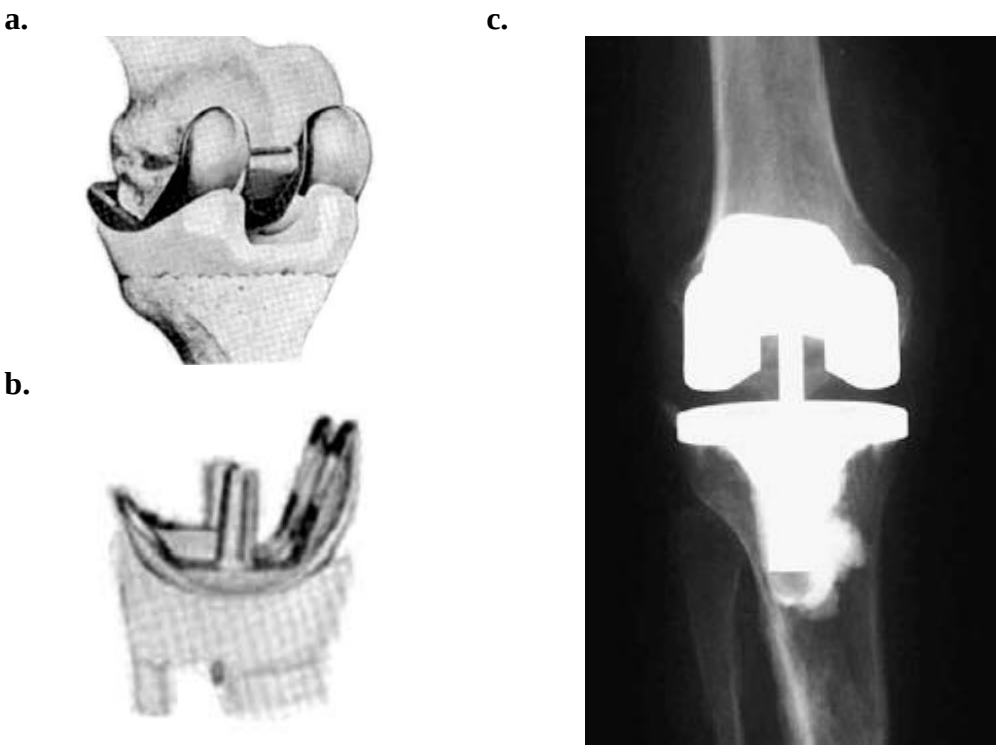


Figure 1.5 Total knee arthroplasty implant **a.** Anterior view **b.** Sagittal view **c.** Radiograph showing constrained total knee implant after surgery (Bonin and Chambat, 2008)

There are 3 main types of implant: constrained, non-constrained and semi-constrained. Constrained implants require large intra-medullary stems which can be the cause of fatigue fractures; however may be useful if there is a large amount of ligament weakness or as a revision surgery. Non-constrained implants possess 5 degrees of freedom (DoF) and conserve the surrounding ligaments. Design of the tibial component must not damage the insertion of the ACL as subsequent joint stability depends on the ligaments rather than the anchoring of the implant. This type of implant may be appropriate for knees where degeneration is not too advanced; however, the ACL may be absent in a large number of OA cases making it unsuitable. Semi-constrained implants are the most common in Europe and North America and are designed to work without ACL conservation. Within this classification there are 2 sub classifications: PCL retaining and posterior stabilised, where the PCL is sacrificed. The PCL is generally found intact in OA cases, hence

the desire to retain it. However, there may be a possibility of decreased joint rotation following surgery (Bonin and Chambat, 2008).

1.3 TKA Rehabilitation

There are a number of outcome measures which can be employed to determine functional outcome following TKA surgery and rehabilitation. These can include subjective questionnaires completed by patients, observational assessment methods completed by physiotherapists, manual measurement methods to determine active and passive range of motion (ROM) and gait analysis (Lowe et al., 2007). This section highlights a small number of outcome measures which routinely appeared in the reviewed literature.

A number of clinically validated patient-reported outcome measures (PROMs) regarding quality of life (QoL) and pain and function in the knee are commonly used to assess function pre and post-op for TKA patients (Clement et al., 2013). Included in this list are the Western Ontario and McMaster Universities Osteoarthritis Index (WOMAC), short form (SF-) 12 and SF-36, the Knee Society Score (KSS) and the Oxford Knee Score (OKS). The WOMAC is a multidimensional questionnaire for patients with OA of the hip or knee which contains pain, stiffness and physical function subscales (Bellamy et al., 1988). The SF-12 is a generic measure of a patient's general physical health and mental well-being, and is a shortened version of the SF-36. The KSS is completed by both the patient and physician and provides a measure of pain, mobility and function whereas the OKS is completed by the patient and assesses pain and function only. Evidence has suggested that condition specific scores, such as the KSS or OKS, are more responsive than general health status

measures such as the SF-12 or SF-36 (Ko et al., 2013). Therefore, only the KSS and OKS will be discussed in more detail.

The KSS is often employed to determine functional outcome following knee replacement surgery. The original Knee Society Clinical Rating System was related to functional abilities such as walking and stair climbing before and after TKA. There was no assessment of accompanying radiographs and the reliability, responsiveness and validity of the score have been challenged (Scuderi et al., 2011). As a result, a new score was devised. The score is patient and physician derived and has pre-op and post-op versions. A section of the score is completed by the physician, detailing level of pain when walking on level ground, stairs and inclines. The physician also completes an assessment of alignment, ligament stability and ROM. The patient then records their satisfaction, functional activities and expectations. The new score is broadly applicable across gender, age, activity level and implant type and is a validated and responsive method for assessing objective and subjective outcome measures following UKA or TKA (Scuderi et al., 2011).

The OKS is a 12 item self-completed, patient based outcome score which has little burden on patients and they have little difficulty completing it (Dawson et al., 1998). Currently, each question has 5 categories of response: 1-5 ranging from least to most difficulty or severity and the total score ranges from 12 (best) to 60 (worst). The OKS has previously demonstrated good validity, reliability and sensitivity (Xie et al., 2011) and is currently the PROM of choice to evaluate TKA in England and Wales (Clement et al., 2013). However, as the OKS is a subjective score, there has been some controversy regarding identification of the minimally important clinical difference (MICD). Clement et al., (2013) aimed to identify the MICD in the OKS

and recorded scores pre-op and 1 year post-op for 505 patients. Results showed that the mean improvement for OKS was 15 points and it was concluded that the MICD difference for the OKS was between 4 and 5 points for both pain relief and function. However, Beard et al., (2015) stated that minimally important difference values are about 5 points, and MICD are about 9 points. It was stated that to assess change over time in a single group of patients the MICD should be used. Therefore, it would appear that there is not a specific MICD when using the OKS, however a difference of between 5 and 9 points may be considered important.

Active and passive range of motion (ROM) is also frequently measured before and after TKA surgery with measurements generally including joint flexion and extension (Mizner et al., 2011; Standifird et al., 2014). Accurate measurement is extremely important as sufficient knee joint ROM is required for a number of activities of daily living (Lavernia et al., 2008). Currently, the majority of clinical centres use manual goniometers to measure knee ROM. However, this may not be the most accurate way to measure ROM either actively or passively (Milanese et al., 2014).

Gait analysis is also used to determine level of function and ability to ambulate independently following TKA. Clinicians generally assess walking speed, range of knee flexion and gait symmetry (Jevsevar et al., 1993; Liikavainio et al., 2007; Ouellet and Moffet, 2002).

Currently, the management and care of TKA patients varies within and across various healthcare systems and uncertainty exists about the optimal process of care, including the appropriate amount of rehabilitation (Brander and Stulberg, 2006; Lingard et al., 2000). According to Brander and Stulberg (2006), rehabilitation

protocols should be designed depending on specific information regarding the TKA surgery. For example, the type of fixation, type and extent of bone cuts, whether patellar surfacing was performed and the type and extent of pre-operative misalignment could all determine which type of rehabilitation protocol should be prescribed. Despite controversies in methods of rehabilitation delivery, there is widespread agreement that early post-op rehabilitation is essential to restore mobility, strength and flexibility and reduce pain and the risk of deep vein thrombosis (Brander and Stulberg, 2006).

A major controversy in current literature is whether rehabilitation should take place in the home or in out-patient centres under the supervision of a physiotherapist (Tian et al., 2009). A number of investigations have examined the effect of home based and clinic based rehabilitation programmes.

Kramer et al., (2003) investigated 160 TKA patients, with one group receiving a home-based exercise rehabilitation programme and the other group receiving individual clinic based treatment provided by outpatient physiotherapists. Outcome measures included the Knee Society Score, 6 minute walk test (6MW) and knee flexion ROM. No significant differences were found between groups at 12 or 56 weeks after surgery.

Mockford and Beverland, (2004) also investigated the effect of 9 weeks of outpatient physiotherapy vs no outpatient physiotherapy in 150 TKA patients. The outcome measure was knee ROM and a significant difference was found in favour of the intervention after 3 months. Rajan et al., (2004) investigated whether outpatient physiotherapy provided any benefit in the short, medium or long term. One hundred and two patients were randomised depending on whether they received inpatient

physiotherapy only or inpatient and outpatient physiotherapy. Patients were seen at 3 months, 6 months and 1 year post-op to record the ROM of the affected knee. There was no statistically significant difference between groups at any of the three examinations. However, the outpatient physiotherapy group displayed slightly higher ROM at 6 months and 1 year.

Controversy also exists over what type of rehabilitation post-surgical TKA patients should receive. Lowe et al., (2007) provide support for exercises based on functional activities after discharge rather than traditional home based exercises. Codine et al., (2004) investigated the effect of submaximal isokinetic hamstring strengthening vs no hamstring strengthening in 60 TKA patients. Results demonstrated a significant difference in knee extension ROM after 30 days favouring the intervention.

Frost et al., (2002) assessed the feasibility of comparing traditional exercise regimes with a more functional and dynamic approach for patients following TKA surgery. Patients were divided into a functional exercise group and a traditional exercise group with both exercise programmes being completed in the home setting. Results at one year follow up demonstrated no statistically significant difference between groups for leg extensor power, walking speed, pain during walking or knee flexion. However, a trend suggested the loss of ROM was slightly less in the functional exercise group.

Moffet et al., (2004) evaluated effectiveness of an intense functional rehabilitation (IFR) programme between the 2^{nd} and 4^{th} months after surgery on functional ability and quality of life (QoL) in TKA patients. The primary outcome measure was the 6MW and secondary outcome measures included functional ability and QoL. The IFR group received a supervised, outpatient rehabilitation programme between

months 2 and 4 following TKA while the control group (CTL) received only a few supervised post-op sessions which focused on traditional rehabilitation movements. Results demonstrated that for the 6MW, the IFR group walked a significantly longer distance than CTL group at 4, 6 and 8 months following surgery. Further, QoL surveys indicated that a significant difference was found in favour of the IFR group in the physical and mental components and the IFR group had less pain, stiffness and difficulty performing activities.

The reviewed literature suggests that the ultimate goal of TKA surgery is to reduce pain, improve stability and restore function to the affected knee. However, evidence suggests functional outcome following TKA is often not restored, with the majority of TKA patients exhibiting lower functional outcome scores than their healthy counterparts (Gill and Joshi, 2001; Noble et al., 2005). It is currently not known if these deficits are the result of disease and surgery or if improvement may be possible with targeted rehabilitation (Negus et al., 2015). A number of studies have investigated functional outcome following TKA. However, comparison of results is difficult as outcome measures and follow-up times often differ between studies. McClelland et al., (2007) conducted a review of studies which used instrumented gait analysis (GA) to assess functional outcome following TKA. A variety of GA specific outcome measures were reported, however the following were consistent amongst most studies: walking speed, knee ROM during gait, maximum knee flexion during swing, maximum knee flexion during stance, range of flexion during loading and sagittal knee moment pattern. For all parameters, at least half of TKA patients exhibited lower scores than controls, indicating a reduced functional outcome. All

reviewed studies reported improvements in patients' gait towards normal, but none reported functional outcomes which matched controls.

Ouellet and Moffet (2002) investigated stair climbing and gait before and 2 months after TKA. Both before and after surgery, speed, cadence and stride length were significantly decreased compared to controls. In particular, the speed deficit was significantly larger 2 months after surgery than before surgery. Further, TKA patients exhibited reduced knee flexion and extension during gait compared to controls at pre-op and 2 months post-op (Figure 1.6).

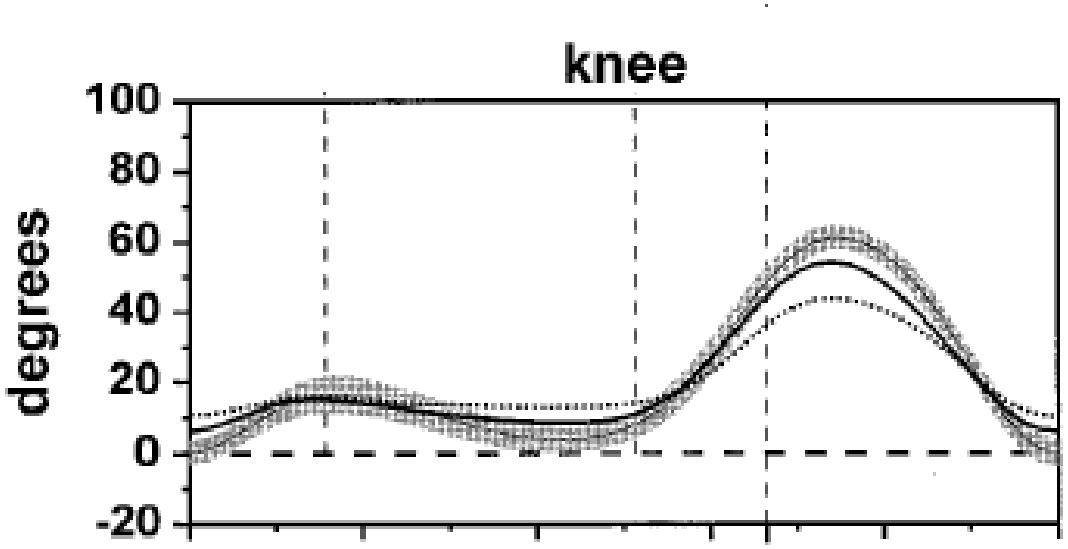


Figure 1.6 Movement at the knee of OA patients before surgery (black), after surgery (dashed) and healthy controls (shaded grey) (Ouellet and Moffet, 2002)

During stair ascent, the cycle duration was significantly increased and cadence significantly decreased compared to controls. A number of large gait and stair ascent deficits were present 2 months after TKA and gait parameters were still below those of pre-op. However, this is a relatively short follow-up time which could contribute to the large deficits. Further, it is stated that patients received 7-10 days of intensive physiotherapy in hospital and were then discharged. Following discharge, very little

physiotherapy was received. Therefore, the gait and stair ascent deficits could be due to lack of appropriate and structured rehabilitation.

Standifird et al., (2014) conducted a review of studies which investigated biomechanics during stair ascent and descent following TKA. Stair ascent was affected more than descent with TKA patients exhibiting reductions in knee flexion at contact, maximum knee flexion, total knee flexion ROM and ascent velocity in comparison to healthy controls. All parameters demonstrated deficits regardless of the implant type, surgeon or staircase design. However, it was stated that studies with long follow-up periods (62-134 months) reported no difference in kinematic or kinetic parameters between TKA and control groups. While this advocates the long term benefits of TKA surgery, short term functional outcomes are still less than ideal.

Milner (2009) conducted a review which aimed to determine whether the biomechanics of gait are abnormal after primary TKA. Six studies investigated peak knee flexion during loading and reported values ranging from 9.8° to 16° for TKA patients and from 16° to 19.7° for controls. The difference between groups was significant for only one study.

This literature review highlights the vast amount of controversy and variability that exists across clinical centres with regards to TKA rehabilitation. It would appear that consolidation of findings is required in order to maximise efficiency of rehabilitation programmes and functional outcome following completion of such programmes. From the reviewed literature it may be suggested that current rehabilitation programmes result in early functional outcomes which are not comparable to healthy counterparts. It may also be proposed that outpatient rehabilitation which focuses on functional movements, aimed at restoring normative gait, may lead to better

functional outcomes than home-based programmes which focus on standard repetitive exercises. Further, there is a lack of direct feedback to patients and clinicians regarding progression through rehabilitation programmes and about the quality of movement whilst completing rehabilitation exercises. As mentioned previously, the most effective way of delivering accurate and reliable feedback to patients is through the use of motion capture technology. However, routine use of motion capture in the clinical environment remains limited.

Therefore, in order to increase the use of motion analysis in routine clinical practice to deliver appropriate feedback and rehabilitation, there is a need for a motion analysis protocol which requires minimal technical expertise to operate and is less time consuming than current systems. This would allow real-time feedback of biomechanical data whilst also providing a stimulating environment for rehabilitation exercises.

1.4 Aims

The primary aim of this PhD was to develop such a tool to augment the rehabilitation experience of patients and clinicians in a realistic clinical environment. Therefore, the first aim was to develop a motion analysis protocol which was tailored for routine clinical use. The second aim was to use the protocol to implement real-time feedback during rehabilitation for TKA patients and the third aim was to test the effectiveness of this type of feedback in a realistic clinical environment.

It was hypothesised that the use of motion capture systems in the clinical environment will be more feasible with the use of a tool which has been designed for this purpose. It was also hypothesised that the reduced complexity of the tool combined with visualisation feedback of movement would lead to a quick and

reliable method for quantifiably measuring and documenting patients' rehabilitation exercises and baseline and functional outcome assessments. With the introduction of such a tool into the clinical environment, the possibility exists to provide accurate and quantifiable movement training and assessments. It was further hypothesised that visualisation feedback of movement to patients will lead to improved functional outcomes, increased patient satisfaction and improved patient clinician dialogue.

Chapter 2: Overview of Measuring Movement in the Clinical Environment

Chapter 2

2 Overview of Measuring Movement in the Clinical Environment

2.1 Introduction to Movement Analysis Methods

In the literature review, mention is made of the fact that the majority of measurement and assessment in current clinical rehabilitation programmes is carried out using observational methods which is unlikely to be the most effective way of assessing patient function and progress. A number of commercially available systems exist which allow more accurate measurement of human movement in 2 or 3 planes which can range from video analysis to fully instrumented 3D analysis. In the following section, various types of movement analysis are introduced along with the advantages and disadvantages accompanying each method. Hardware and software configurations used throughout the study are detailed, along with definitions for global, anatomical and technical reference frames.

As mentioned previously, the majority of gait and functional movement assessments are still based on observational analysis (Toro et al., 2003). A number of validated scoring methods exist in an attempt to quantify gait assessment by observational analysis (Mackey et al., 2003; Read et al., 2003). However, it has been suggested that these may have high inter and intra-assessor variability and are therefore unlikely to be a suitable replacement for 3D motion analysis (Kawamura et al., 2007; Ong et al., 2008). Further, the majority of observational scoring methods have been developed specifically for measurement of gait and therefore are not appropriate for quantifying functional movement or ROM during other functional tasks. Currently, the majority of physiotherapists use manual goniometers to measure static active and passive joint ROM at baseline and outcome assessments. While this method is more

accurate than observational methods, there is still significant variability when compared to instrumented measurement methods (Nussbaumer et al., 2010), and further, it cannot be used to measure dynamic functional movements.

One method of augmenting observational analysis is through the use of 2D video footage. A number of software packages are available which are capable of measuring and quantifying gait and functional movements from 2D video footage (Contemplas, 2015; Simi, 2015). However, these movements can only be considered valid if the movement is in the plane of the camera and the camera is perpendicular to the subject. Further, these packages generally require the user to identify joint centres in the sagittal or coronal planes, thus allowing the software to calculate joint angles in one plane using simple geometry. However, this method is still subject to user variability in the identification of joint centres. Further, users will often be required to process a gait or functional movement trial on a frame by frame basis, making the process time consuming and also increasing joint centre location variability. More recent software iterations may employ pixel recognition techniques to semi-automate the analysis process although the method still remains limited to one plane of movement square on to the camera. Additionally, very few of these packages have been validated for clinical use, thus widespread use in the clinic is limited.

2.1.1 Three Dimensional Analysis

Since the commercialisation of 3D motion capture technology, it has become the gold-standard for measurement and assessment of human movement. Throughout this study, 3D motion analysis will be referred to as instrumented motion analysis. Instrumented motion analysis allows objective and accurate measurement of

movement in 3 planes. Figure 2.1 is an overview of the components required for an effective instrumented motion analysis session. The hardware and software components will be discussed in the following section and the model will be discussed in chapter 3.

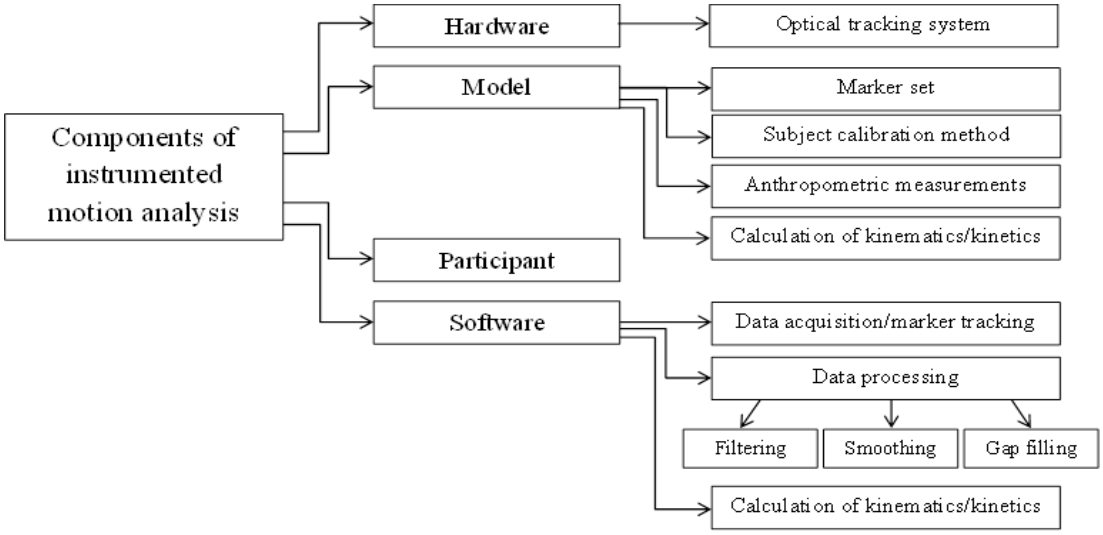


Figure 2.1 Components required for an instrumented motion analysis session

In terms of hardware, there are a number of commercially available instrumented motion capture systems which have been validated for clinical use (Barker et al., 2006). These systems use infra-red light emitting cameras which track active or passive markers located at specific locations on a participant's body allowing calculation of a number of biomechanical parameters. Passive markers are retroreflective, allowing them to reflect infra-red light emitted by the cameras. Other surfaces such as skin, clothes, flooring and furniture do not usually readily reflect infra-red light and hence the markers appear as a light dot on a dark background and are therefore easy to detect. Conversely, active markers emit infra-red light, which is tracked by the cameras. Active markers can be useful when measuring over long distances or large volumes (O'Nolan, 2013); however, they tend to be more expensive than passive markers and require a battery or power source which may

impede the participant's movement. In the majority of cases, passive markers are a suitable alternative to active markers as they are much cheaper and do not require a power source.

Cameras used to track markers can vary in cost and accuracy depending on the system. Generally, cameras will capture at speeds of 100 to 120 frames per second (fps) which ensures capture of discrete human movements and events, even at high speeds. Across a range of commercially available systems, the mean measurement error of tracked markers can range from 0.1mm to 6.0mm depending on the age of the system, number of cameras and specification of cameras (Chiari et al., 2005). A number of systems will also allow camera properties such as exposure and aperture to be adjusted manually, allowing a customisable setup for different motion capture scenarios and capture volumes.

The majority of commercially available motion capture systems come with their own software package which is tailored to work with specific hardware. The software can be the most powerful tool in the motion analysis process, depending on how many options for processing and analysis it offers. Leading commercial motion capture systems offer a vast number of options for processing and analysing data. Automated processes usually involve participant calibration, marker labelling, gap filling during dynamic trials, filtering and smoothing of data and calculation of kinematic and/or kinetic data. These are just a few of the processes that a software package may offer. While it is useful that these processes are automated, it leaves little room for the user to customise certain aspects. For example, if labelling and calculation of kinematics are automatic, the user must apply the appropriate marker set in order to carry out these processes.

The output of data from most commercially available systems can vary, depending on what the user specifies. Most will allow output of an excel file containing any data from marker trajectories to joint angles, forces and moments. It is then up to the user how to interpret and display the information. While most commercial motion capture software platforms provide some level of reporting, the extraction and interpretation of specific data can be a very complex and time consuming process and even then, displaying the data can often be limited to standard data processing package functionalities.

2.2 Typical Laboratory Configuration

This study employed an 8 camera Vicon Bonita (Vicon Motion Systems, Oxford, UK) optical tracking system which is less expensive than the current state of the art camera systems while still maintaining high levels of accuracy. The manufacturer's website claims that the Bonita can capture with precision down to 0.5 mm of translation. Further, the Bonita can capture at up to 250 fps, which is ample frequency for capturing discrete movement events.

Prior to each session, the system was calibrated using a precision engineered calibration wand manufactured by Vicon (Figure 2.2). Five active markers are present on the wand and the distance between each marker is known by the system. By measuring the appearance of the markers on the wand in the field of view of each camera, the software can accurately estimate the position of each camera relative to the wand and each other. This process is called camera calibration and it allows definition of the capture volume. The wand is then placed on the floor and a further capture is undertaken to define the ground plane and axes of the global reference frame (GRF; Figure 2.2b).

a. **b.**

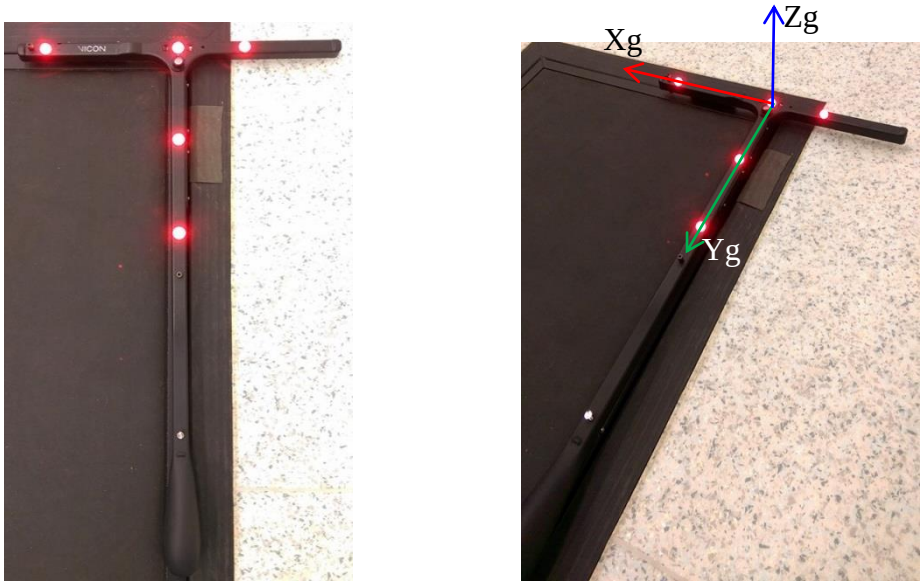


Figure 2.2 Wand used to define the global reference frame **a.** Transverse view **b.** 3D perspective view with global reference frame axes as defined by the wand

Calibration returns a residual measurement value for each camera which provides information about the accuracy of the 3D measurement of a marker. Residual values measure the distance from the ray of the camera to the assumed location of a marker (Figure 2.3).

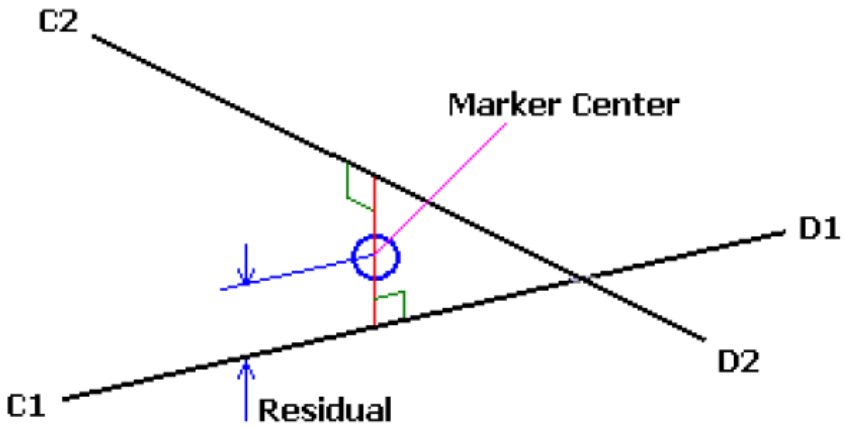


Figure 2.3 Marker centre residual determination with 2 cameras (Motion Lab Systems, 2016)

In reference to Figure 2.3; camera 1 and camera 2 both observe the marker centre to be on lines D1 and D2, respectively. Therefore, it can be assumed that the marker must lie on the point where these lines intersect. However, in modern motion

capture more than 2 cameras are often required and with multiple cameras, camera rays don't necessarily intersect (Figure 2.4), so this method cannot be relied upon to determine the position of the marker centre. Therefore, the software must make a decision about the most probable location of the point. The distances from the assumed marker location to each ray are related to the uncertainty of the marker's calculated location and are termed residuals. The Vicon software implements a "least-squares" method (Figure 2.4) which calculates the location of the marker centre such that the sum of the squares of the shortest distances from that point to each ray is a minimum (Motion Lab Systems, 2016).

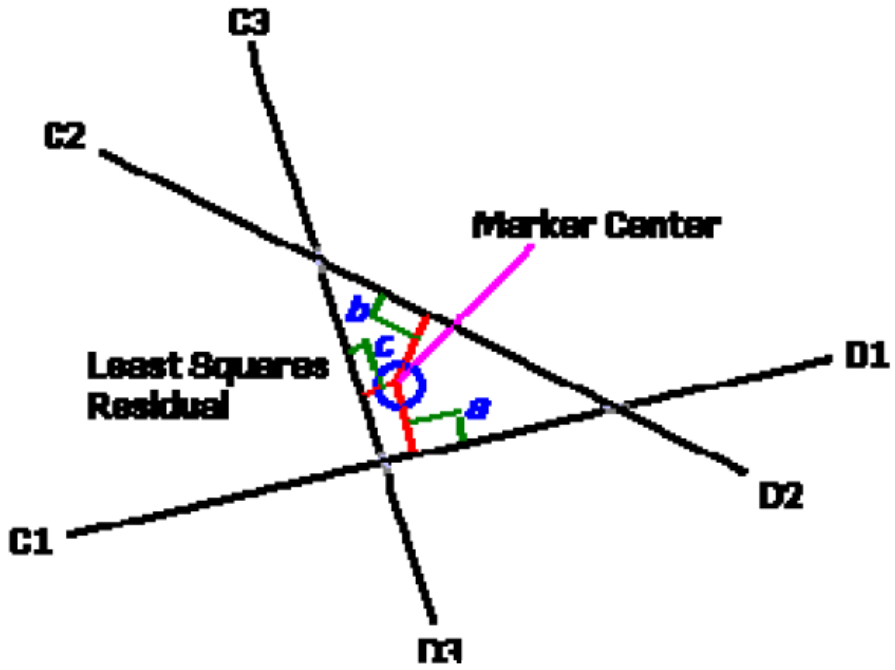


Figure 2.4 Marker centre residual determination with 3 cameras, implementing a "least-squares" method (Motion Lab Systems, 2016).

These distances are the residuals for each camera and the smaller a residual, the more accurately the camera is measuring the true location of the marker centre. System calibrations are usually deemed successful if each camera returns a mean residual of 0.5 mm or less.

The wand in Figure 2.2 was also used to define the lab ground plane and determine the origin and direction of the global x, y and z axes which comprise the GRF. This will be discussed in more detail in chapter 3.

2.3 Software Configuration

Two main software platforms were used in this study. Marker trajectories were identified using Vicon Tracker 3.2.0 (Vicon Motion Systems, Oxford, UK). Marker positions were defined within the GRF such that each marker had an x, y and z position describing its 3D movement.

Marker trajectories were then streamed into D-Flow (Motekforce Link, Netherlands) which is a module based application development package capable of receiving marker trajectory data along with a number of other input options such as force transducers or switches. Input data can be manipulated using a number of available modules or by using script written in Lua code. Further, a number of inbuilt objects are available which allow simple object based programming for development of visualisations.

The appropriate biomechanical models were then applied in order to produce a kinematic output. The maths behind the biomechanical model is described in detail in chapter 3. Object orientated programming was used to develop an avatar and visualisations for a variety of rehabilitation exercises. These methods are discussed in more detail in chapter 6.

2.4 Reference Frame Definitions

In this study there are 3 reference frames of relevance. Initially, the GRF allows description of points in 3D space. Within the GRF, the technical reference frame

(TRF) describes segment movement and the anatomical reference frame (ARF) describes bone movement in reference to the TRF (Figure 2.5).

The TRF can be defined in different ways depending on the markers used to define it. The only requirement for definition of a TRF is at least 3 non/collinear markers on the segment of interest. The markers can either be individual markers, attached to the surface of the skin, or a cluster of markers arranged on a rigid plate. This study employed cluster markers, a schematic example of which is shown in Figure 2.5. The use of single skin surface markers and clusters of markers is discussed in more detail in chapter 3.

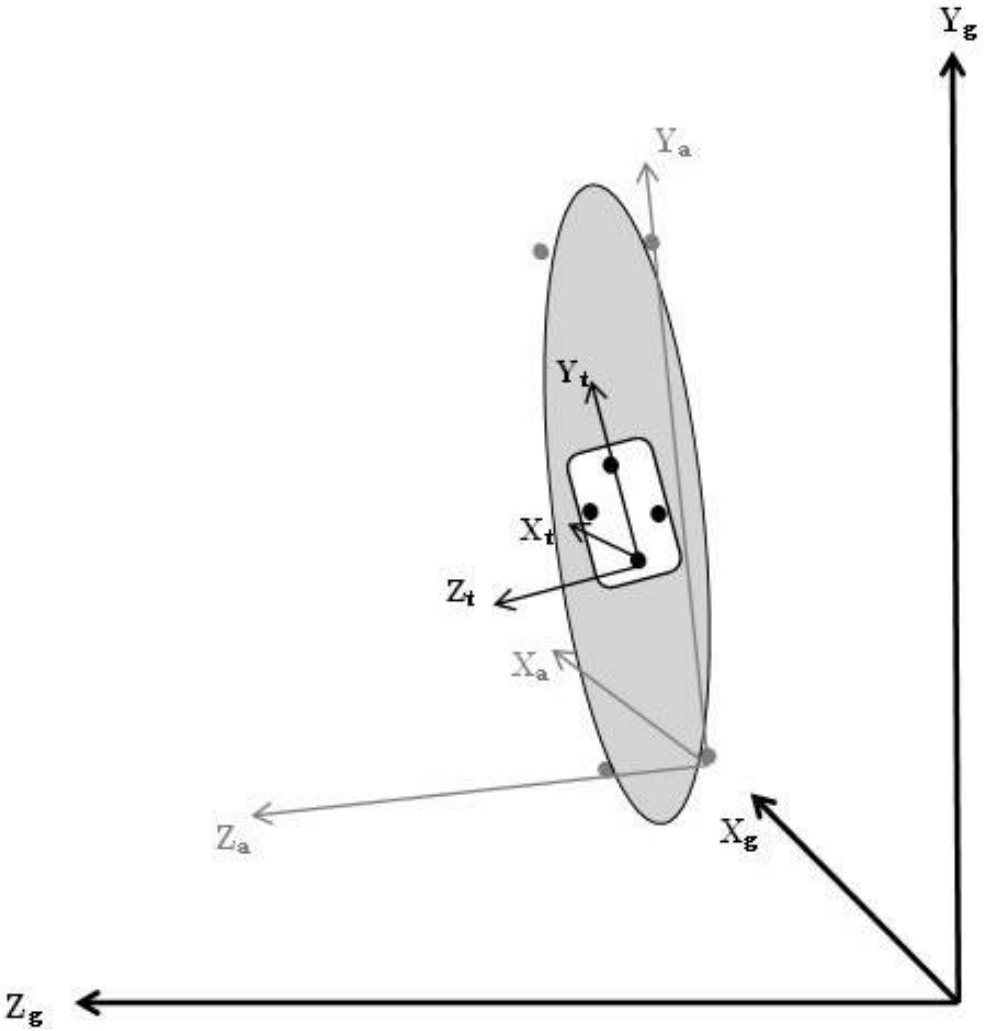


Figure 2.5 Representation of global, technical and anatomical reference frames for one segment. Subscript g indicates axes associated with the global reference frame. Subscript t indicates axes associated with the technical reference frame. Subscript a indicates axes associated with the anatomical reference frame

Technical and anatomical reference frames are calculated using Equation 2.1 through Equation 2.4 (appendix 1). Where A_1 is the first axis, A_T is a temporary axis, A_2 is the second axis, A_3 is the third axis and $P_1 - P_4$ are the four markers associated with the respective segment.

$$\begin{bmatrix} x \\ y \\ z \end{bmatrix}_{A_1} = \frac{\left(\begin{bmatrix} x \\ y \\ z \end{bmatrix}_{P_1} - \begin{bmatrix} x \\ y \\ z \end{bmatrix}_{P_2} \right)}{\left| \begin{bmatrix} x \\ y \\ z \end{bmatrix}_{P_1} - \begin{bmatrix} x \\ y \\ z \end{bmatrix}_{P_2} \right|}$$

Equation 2.1

$$\begin{bmatrix} x \\ y \\ z \end{bmatrix}_{A_T} = \frac{\left(\begin{bmatrix} x \\ y \\ z \end{bmatrix}_{P_3} - \begin{bmatrix} x \\ y \\ z \end{bmatrix}_{P_4} \right)}{\left| \begin{bmatrix} x \\ y \\ z \end{bmatrix}_{P_3} - \begin{bmatrix} x \\ y \\ z \end{bmatrix}_{P_4} \right|}$$

Equation 2.2

$$\begin{bmatrix} x \\ y \\ z \end{bmatrix}_{A_2} = \frac{\left(\begin{bmatrix} x \\ y \\ z \end{bmatrix}_{A_1} \times \begin{bmatrix} x \\ y \\ z \end{bmatrix}_{A_T} \right)}{\left| \begin{bmatrix} x \\ y \\ z \end{bmatrix}_{A_1} - \begin{bmatrix} x \\ y \\ z \end{bmatrix}_{A_T} \right|}$$

Equation 2.3

$$\begin{bmatrix} x \\ y \\ z \end{bmatrix}_{A_3} = \frac{\left(\begin{bmatrix} x \\ y \\ z \end{bmatrix}_{A_1} \times \begin{bmatrix} x \\ y \\ z \end{bmatrix}_{A_2} \right)}{\left| \begin{bmatrix} x \\ y \\ z \end{bmatrix}_{A_1} - \begin{bmatrix} x \\ y \\ z \end{bmatrix}_{A_2} \right|}$$

Equation 2.4

This results in a 3x3 rotation matrix containing the direction cosines for each axis of the orthogonal axis system which makes up the reference frame. These matrices allow points to be transformed between reference frames. This process will be discussed in more detail in chapter 3.

2.4.1 Global Reference Frame

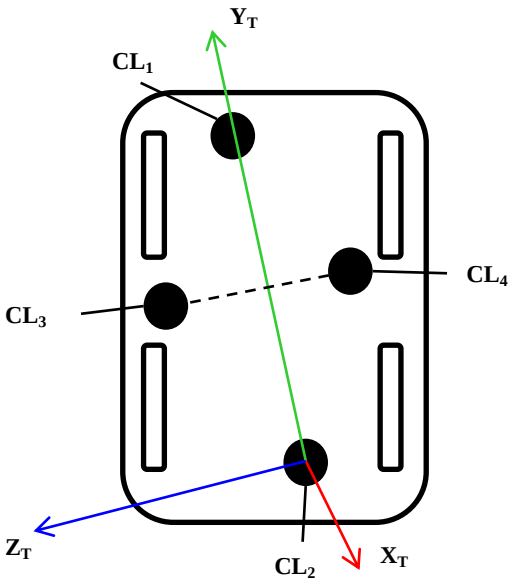
To describe points in 3D space, it is necessary to define the GRF within the capture volume. As mentioned previously, this is usually done using a precision engineered, rigid wand or frame thus allowing definition of the x, y and z axes (Figure 2.2). The wand is placed on the ground in the same position each time. In this study, the corner of a pressure plate was used in order to ensure the wand was placed in the same position during each calibration (Figure 2.2). A spirit level is also present in the wand, to ensure that definition of the ground plane is truly in line with the ground.

Once these axes have been defined, any marker within the capture volume can be described in terms of its x, y and z position. It should be noted that the GRF axes in Tracker differ from the GRF axes in D-Flow. In Tracker, Z is up whereas in D-Flow, Y is up. However, D-Flow accounts for this difference and therefore correction of marker trajectories between software platforms is not required.

2.4.2 Technical Reference Frames

This study implemented a cluster based marker set, the development and design of which will be discussed in more detail in chapter 3. Each segment requires a rigid plate with 4 markers which are used to estimate the position and orientation of the segment during static and dynamic movement trials. This is done by creating a technical reference frame (TRF) for each cluster. The TRF is described in the same way for all clusters and is described in Table 2.1.

Table 2.1 Technical reference frame definition

	Origin CL_2 X_T Mutually perpendicular to Y_T and Z_T Y_T Line between CL_1 and CL_2 Z_T Line between CL_3 and CL_4 and mutually perpendicular to X_T and Y_T

Upon calculation of each TRF a 3x3 rotation matrix is derived which can be used to transform a point from the GRF to the TRF and vice versa. The TRF allows description of segment movement within the GRF, but in order to calculate accurate kinematics, description of segments relative to one another is necessary. This requires definition of anatomical reference frames (ARFs) for each segment which can then be described relative to the TRF.

2.4.3 Anatomical Reference Frames

The lower body comprises 7 segments which are assumed to be rigid; the pelvis, left and right thighs, left and right shanks and left and right feet. Anatomical frames are defined using palpable anatomical landmarks associated with each segment. Anatomical landmarks must be suitable so that internal joint centres can be estimated. For lower limb evaluation, the hip joint centre (HJC), knee joint centre (KJC) and ankle joint centre (AJC) must be defined and anatomical landmarks must also be suitably located to allow the definition of anatomical reference frames. Anatomical landmarks used in this study include the left and right anterior superior

iliac spines (LASIS; RASIS), left and right posterior superior iliac spines (LPSIS; RPSIS), left and right medial and lateral epicondyles (LME; LLE; RME; RLE), left and right medial and lateral malleoli (LMM; LLM; RMM; RLM), the left and right calcaneus (RCA; LCA), the left and right head of the first and fifth metatarsals (LFM; LVM; RFM; RVM) and the apex of the big toe (Toe; Figure 2.6).

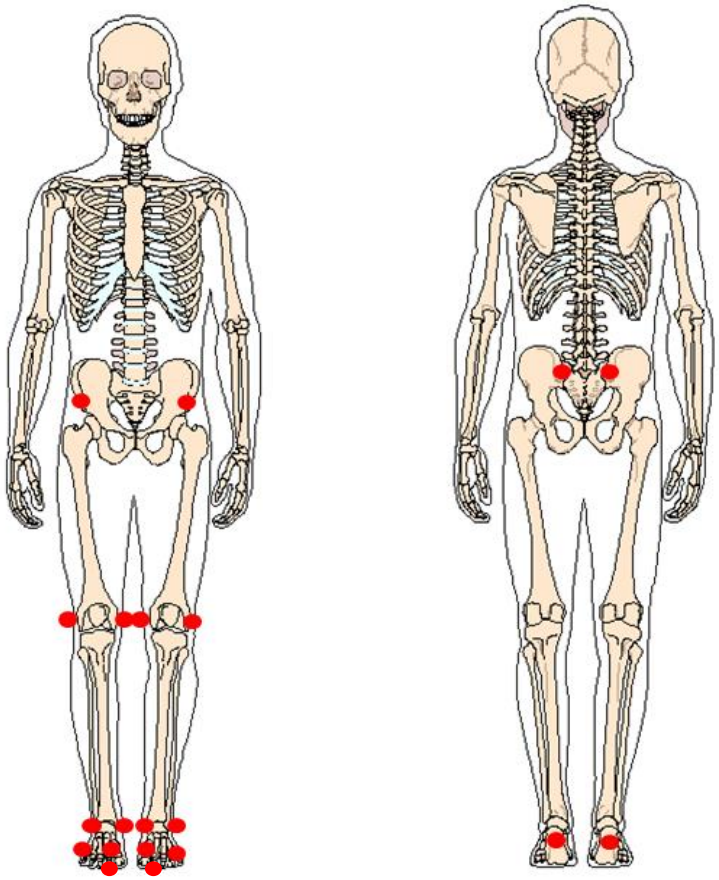
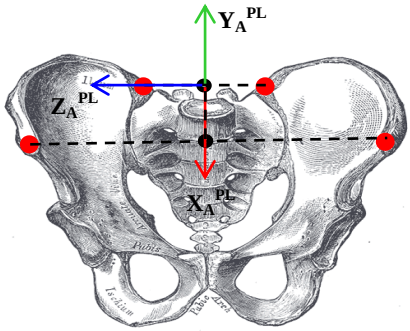
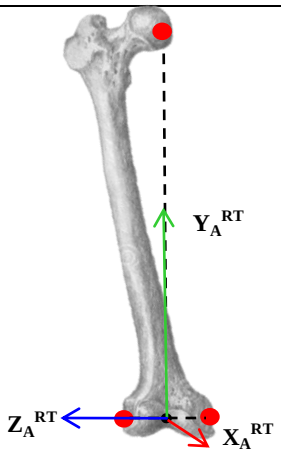
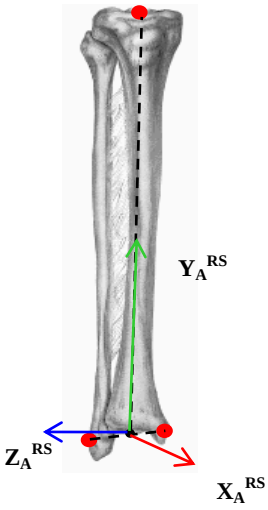
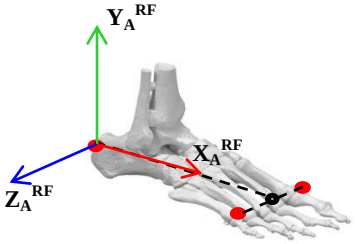


Figure 2.6 Anatomical landmarks for estimation of joint centres and anatomical reference frame definitions

In this study all anatomical reference frames except the foot are calculated in accordance with the International Society of Biomechanics recommendations (Grood and Suntay, 1983; Wu et al., 2002; appendix 1) and they all comprise an orthogonal, right-handed axis system. International recommendations suggest the foot be modelled as a single vector and therefore only ankle plantar/dorsi flexion and

internal/external rotation can be reliably measured. This study modelled the foot as a 3D segment and therefore more reliable measures of foot inversion/eversion could also be obtained. Anatomical frame definitions for each segment are described in Table 2.2. Definitions for right segments are provided. Left segment ARFs are defined in the same way, with the Z axis always being positive to the right.

Table 2.2 Anatomical reference frame definitions for the pelvis, right thigh, right shank and right foot

	Origin	Midpoint between the LPSIS and RPSIS
	X_A^{PL}	Line between the midpoint of the RASIS and LASIS and the midpoint of the RPSIS and LPSIS
	Y_A^{PL}	Mutually perpendicular to X_A^{PL} and Z_A^{PL}
	Z_A^{PL}	Mutually perpendicular to X_A^{PL} and Y_A^{PL}
	Origin	KJC (midpoint between RME and RLE)
	X_A^{RT}	Mutually perpendicular to Y_A^{RT} and Z_A^{RT}
	Y_A^{RT}	Line between HJC and KJC
	Z_A^{RT}	Mutually perpendicular to Y_A^{RT} and X_A^{RT}
	Origin	AJC (midpoint between RMM and RLM)
	X_A^{RS}	Mutually perpendicular to Y_A^{RS} and Z_A^{RS}
	Y_A^{RS}	Line between AJC and KJC
	Z_A^{RS}	Mutually perpendicular to X_A^{RS} and Y_A^{RS}
	Origin	CA
	X_A^{RF}	Line between CA and the midpoint between FM and VM
	Y_A^{RF}	Mutually perpendicular to X_A^{FT} and Z_A^{FT}
	Z_A^{RF}	Mutually perpendicular to X_A^{FT} and Y_A^{FT}

Upon calculation of each ARF, a 3x3 rotation matrix is derived which details the direction cosines of the 3 axes.

Once all reference frames have been defined, the matrices produced from calculation of the TRF and ARF can be used to transform a point from the GRF to the ARF. More specifically, anatomical landmarks introduced in Figure 2.6 can be described within the TRF, with respect to movement of the cluster. Once anatomical landmarks have been described in the TRF, they can be used to define the ARF. Therefore, matrix transformation from one reference frame to another, using the matrices derived in this section, allow the ARF to be described with respect to the TRF and bone movement to be described relative to movement of the cluster attached to the segment. The specific maths behind this process will be discussed in more detail in chapter 3.

Chapter 3: Development of a Bespoke Biomechanical Model

Chapter 3

3 Development of a Bespoke Biomechanical Model

3.1 Introduction

In this chapter the principles of gait analysis are introduced and the purpose of a biomechanical model is outlined, along with details of the components which comprise a biomechanical model. A short literature review of comparisons between different biomechanical models and their kinematic outputs is provided. The development of the bespoke model used in this study is detailed and the methodology and results for a comparison to the current clinical gold standard model is described.

3.2 The Gait Cycle

In human movement science, gait is often assessed as the ability to ambulate independently and efficiently and can often be a priority outcome for a number of treatment interventions (McClelland et al., 2007). Gait analysis (GA) is an extremely important tool which allows assessment of a number of different gait parameters at different stages of recovery or treatment and can indicate how a patient is progressing towards efficient ambulation. Gait is often described in cycles (Figure 3.1); with a full cycle described as initial contact of one foot to subsequent contact of the ipsilateral foot (Baker, 2013). The cycle is usually divided into phases to more clearly describe the events occurring. The simplest division is to separate the cycle into stance phase, when the foot is in contact with the ground, and swing phase, when it is not. Stance begins with initial contact (IC) of the foot with the ground which is shortly followed by loading response (LR) as the limb bears the weight of the body and the ground reaction vector (GRV) is generated (Gage et al., 2009).

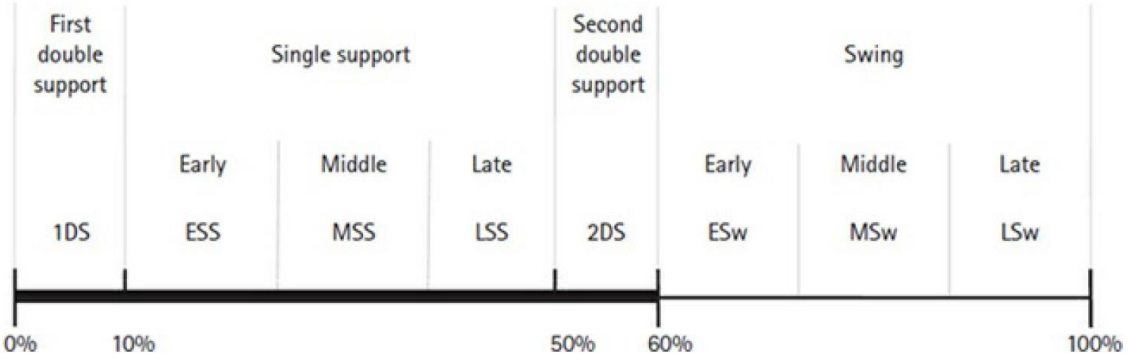


Figure 3.1 Stages of the gait cycle (Baker, 2013)

Stance phase is often further subdivided into first double support, when both feet are initially in contact with the ground, single support, when only the ipsilateral foot is in contact with the ground and second double support, when the contralateral foot is back in contact with the ground (Baker, 2013). Stance phase has also been subdivided into mid stance, terminal stance and pre-swing (Gage et al., 2009). The point at which the foot is no longer in contact with the ground and stance phase ends is termed 'foot-off'.

Swing phase occurs immediately after stance. It begins when the foot is no longer in contact with the ground and ends when it comes back into contact with the ground. Swing is often divided into three sub phases of equal duration; initial swing; mid swing and terminal swing (Baker, 2013; Gage et al., 2009). Swing phase occupies approximately 40% of the cycle and a number of important aspects of gait occur during this phase. For example, swing allows forward advancement of the limb by providing foot clearance of the ground. Subconsciously, we select a walking speed which allows the limb to swing with minimal extraneous muscle action, therefore reducing energy expenditure. In this way, swing also helps to conserve energy during gait (Gage et al., 2009). The cycle is often normalised to 100% and certain events are expected to occur at certain time percentages of the cycle in normal

subjects. For example, foot-off is expected to occur at approximately 60% of the gait cycle (Baker, 2013) and deviations from the time at which these events occur may help to indicate an abnormality or impairment.

Inman et al., (1981) stated that there are two basic requisites to the act of walking 1) "Continuing ground reaction forces that support the body" and 2) "Periodic movement of each foot from one position of support to the next in the direction of progression". It was suggested that these basic requisites are necessary for any form of bipedal walking and as long as they are present then forward progression should be possible, no matter how marred by physical disability. Gage et al., (2009) expanded on this idea by defining the "five prerequisites of normal gait". These are as follows:

- Stability in stance
- Sufficient foot clearance in swing
- Appropriate swing phase pre-positioning of the foot
- Adequate step length
- Energy conservation

Stability in stance is achieved by maintaining the centre of mass (CoM) above the base of support. During walking, body segment alignment is constantly changing which results in a moving CoM. To compensate for this movement the trunk must alter its position to maintain balance and keep the CoM over the base of support. Failing to do so will result in an external moment caused by the misalignment of the CoM which would cause excessive muscle contraction to correct for the moment (meaning that gait is no longer being achieved with minimal energy expenditure) or if the moment was too large to correct, a fall would occur.

Clearance in swing phase is dependent on a number of factors. On the stance side the ankle, knee and hip must be appropriately positioned and be capable of producing the appropriate power. The stance limb is the primary driver for forward motion during normal gait so appropriate power across the joints is essential to ensure adequate forward progression (Gage et al., 2009). On the swing side there must be adequate ankle dorsiflexion, knee flexion and hip flexion allowing the foot to clear the ground. Further, there must be stability of the stance foot and adequate body balance to keep the body upright. The natural reaction would be for the body to list towards the swing side, causing the swinging limb to come into contact with the ground. Therefore, appropriate balance from the stance foot and trunk are required to keep the CoM above the base of support and prevent listing (Inman et al., 1981).

In order to achieve preposition of the foot in terminal swing the stability, power and proper positioning previously mentioned must continue. However, most importantly, the ankle must be able to dorsiflex adequately and there must be appropriate balance between the inverters and everters of the foot (Gage et al., 2009). Inadequate dorsiflexion would result in the subject landing with a flat foot or with toes first instead of heel, which can lead to a number of biomechanical abnormalities (Williams et al., 2014). Similarly, if the foot is not correctly everted so that the lateral side of the plantar surface of the foot comes into contact with the ground first, this can lead to misalignment of the ankle which can also lead to a number of more proximal biomechanical issues (Rao et al., 2012).

An adequate step length requires a combination of the factors discussed, such as sufficient body balance, proper positioning on the stance side, adequate joint flexion or extension at the hip and knee on the swing side and proper foot positioning.

Without these factors there is likely to be instability or insufficient room for appropriate swing, leading to premature foot contact and ultimately an inadequate step length.

Finally, energy conservation requires the appropriate line of action of the GRV, thus allowing passive joint stability. If the magnitude or direction of the GRV is out with normal parameters, muscle action will be required to counteract the internal moments. At each stage of the cycle the GRV generates external moments which allow passive support, thus limiting the need for muscle action and generation of internal moments (Figure 3.2; NHS Scotland, 2009). This allows the stance limb to support the weight of the body and facilitate forwards progression with minimal active muscle contraction.

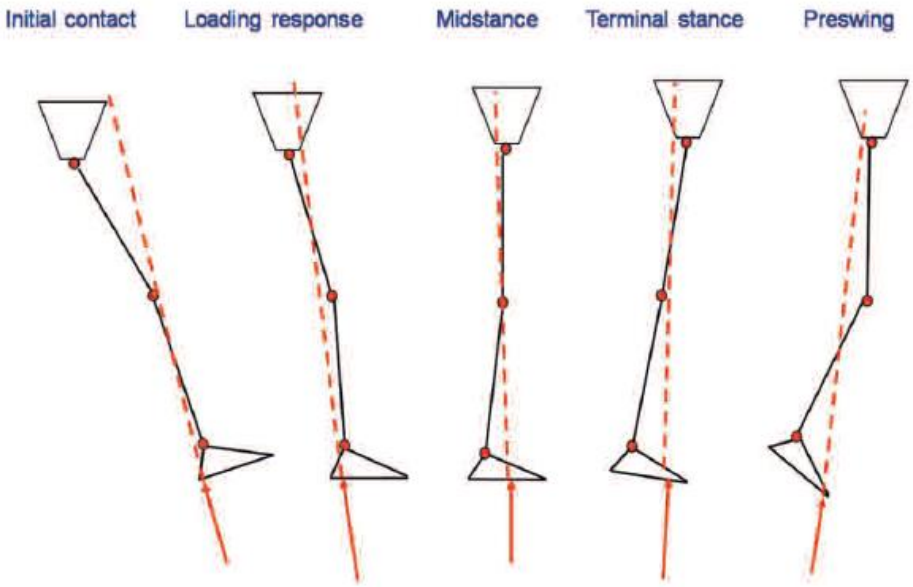


Figure 3.2 Normal progression of the GRV (NHS Scotland, 2009)

3.3 Spatiotemporal Parameters of Gait

Spatiotemporal parameters are often used in observational analysis of pathological gait by comparing the subject's spatiotemporal outputs to that of a non-pathological

individual to determine if an abnormality may be present. Commonly measured spatial parameters include step length, step width and stride length. One step is defined as the movement of one foot over the other and step length is defined as the distance one part of the foot travels in front of the same part of the other foot during each step. A stride is a step for one foot followed by a step for the other (Figure 3.3).

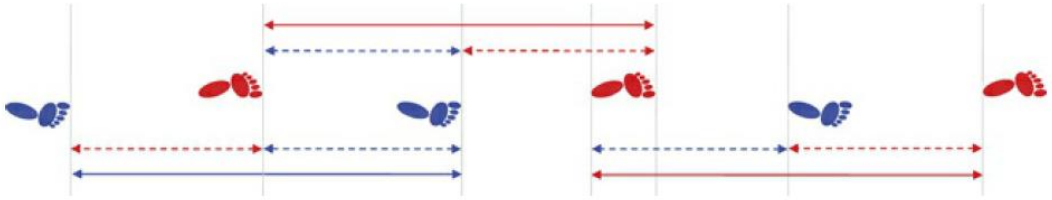


Figure 3.3 Step (dashed) and stride (solid) lengths for symmetrical walking (Baker, 2013)

Step width is often classified as the medio-lateral separation of the feet. Ultimately, this measurement will depend on which part of the foot the measurement is taken from. The distance between the heels is commonly used, however if 3D analysis is available then the distance between the ankle joint centre (AJC) is recommended (Baker, 2013).

Commonly measured temporal parameters include stride time, cadence and walking speed. Stride time is defined as the duration of one gait cycle; however, cadence is more often employed to measure the duration of a gait cycle and is calculated by the number of cycles in a specified time. Walking speed can be a very useful tool in observational analysis of gait as an increase in self-selected walking speed can often indicate functional recovery (Abbasi-Bafghi et al., 2012; Baker, 2013). The 6 minute walk test (6MW) is routinely used to determine if there has been an increase in walking speed over a prescribed time period (Mizner et al., 2011).

3.4 Gait Symmetry

Gait symmetry is defined as perfect agreement between the actions of the lower limbs and may be suggested if no statistical difference is measured between parameters measured bilaterally (Sadeghi et al., 2000). Symmetry is often assumed in healthy individuals for the sake of simplicity in data collection and analysis (Gundersen et al., 1989) and therefore, gait asymmetry is often considered to indicate gait pathology. Gait asymmetry has previously been considered important clinically as it may be associated with inefficiency, challenged balance control, risk of musculoskeletal injury to the dominant side and loss of bone mass in the non-dominant side (Patterson et al., 2010).

However, a number of studies have indicated that symmetry may not be assumed in normal gait. Sadeghi et al., (2000) carried out a review to summarise work regarding the assumption of lower limb symmetry during able bodied gait and reported that asymmetry in able bodied subjects has been described frequently. Asymmetry has been reported in the following parameters: velocity profiles, step and stride length, foot placement angle, max knee flexion and ROM (Gundersen et al., 1989; Patterson et al., 2010; Sadeghi et al., 2000). It is suggested that asymmetry in able bodied subjects may be explained by the actions taken by the lower limbs to propel body segments and control forwards progression. The review also stated that there has been no conclusive study on the influence of lateral dominance on the symmetrical or asymmetrical behaviour during able-bodied gait.

Gundersen et al., (1989) aimed to examine the assumption of symmetry and determine if lower extremity lateral dominance played a role in any asymmetries observed. The following kinematic variables were examined: step time, stance time,

total gait cycle time, step length, max knee flexion/extension, max ankle dorsi/plantar flexion and percentage of the gait cycle at which the latter four stages occurred. Asymmetry was defined as a statistically significant difference between limbs for each variable measured. Fourteen healthy volunteers performed 8-10 walking trials and analysis was carried out in the sagittal plane only. Results suggested that neither symmetry nor asymmetry in the variables could be generalised across subjects and symmetry should not be used in absolute terms when applied to human function. Further, it was concluded that asymmetry cannot be predicted by lateral dominance.

Kobayashi et al., (2014) evaluated gait symmetry in the normal Japanese population and examined the effect of age and gender on gait features by autocorrelation of trunk acceleration. Forty elderly people (mean age 70) and 47 university students (mean age 20) volunteered for the study. Each performed 4, 7m walks with an accelerometer attached to the lower back and an autocorrelation function was used to calculate symmetry. Results suggested that age and gender had a significant effect of gait symmetry with older people being less symmetrical and females being more symmetrical. It was also noted that gender differences tended to be larger in elderly populations.

Rapp et al., (2015) aimed to determine changes in symmetry during the rehabilitation process of patients following total hip arthroplasty (THA) using inertial sensor technology and autocorrelation analysis. Twenty-nine patients and 30 age-matched, healthy controls were recruited. Symmetry was assessed using gait tested on 3 occasions with at least 6 days between sessions. Patients' asymmetries decreased

over an average of 27 days of inpatient rehabilitation however symmetry did not reach the values of the control group, even after intensive rehabilitation.

From these studies it can be determined that symmetry cannot be assumed in normal individuals and lateral dominance does not appear to play a role in any asymmetries which may be present. However, it may be suggested that individuals with a pathology may exhibit higher levels of asymmetry than healthy individuals. Further, there doesn't appear to be a universally accepted method for measuring and quantifying symmetry as the studies discussed above implement different methods and use different parameters to calculate symmetry. As a result of this, a number of authors have investigated the use of different parameters or indices to calculate symmetry.

One of the most commonly used indices is the symmetry index (SI; Table 3.1) which was developed by Robinson et al., (1987). It is assumed that when SI = 0, gait is symmetrical. However, there are a number of limitations with this index. Differences are reported against their average value and parameters that have large values but relatively small inter-limb differences tend to lower the index and reflect symmetry (Sadeghi et al., 2000).

Patterson et al., (2010) aimed to compare the properties of commonly used expressions of gait symmetry with the goal of achieving a recommendation for standard practice. Further, they aimed to determine if different symmetry measures provided unique information about characteristics of gait symmetry in a group of community-dwelling, ambulant individuals post-stroke. Focus was placed on spatiotemporal parameters (SPTs) as these have been most commonly used in the past. Five gait parameters and 4 equations were analysed. The parameters were step

length, stance time, swing time, double support time and intra-limb ratio of swing/stance time and the equations were symmetry index (SI), symmetry ratio (SR), log-transformed symmetry ratio (GAsym) and symmetry angle (SA; Table 3.1). One-hundred and sixty one stroke patients were recruited along with 81 healthy subjects to establish confidence intervals for 'normal' symmetry and 3 over ground walking trials were performed at a self-selected pace. Results suggested that individual equations did not provide any unique differences; however altering the gait parameter used in the symmetry equation may have an effect on the result. It was also suggested that the ratio equation may be easier to interpret. For example, an individual with a swing ratio of 2.0 has an affected side swing duration twice as long as the non-affected side. It was also recommended that the numerator should always be the greater of the 2 values, regardless of the affected side and information about the direction of symmetry could be retained with a sign convention. Double support time symmetry identified the fewest asymmetric individuals and therefore was not recommended to distinguish individuals within this population. Step length, swing time and stance time were identified as the most useful parameters.

Błażkiewicz et al., (2014) aimed to determine which methods resulted in the highest diagnostic values in relation to kinematic data. Currently, the most common approaches are the SI and RI. Disadvantages of these methods include the potential for artificial inflation, for example if one side is < 0 and one side is > 0, and also that symmetry cannot be evaluated through one complete cycle. Other methods include principle component analyses, regions of deviations analysis and paired t-test; however these may require additional subjects and experiments may need normative data as a reference. Spatiotemporal parameters of 58 able-bodied participants were

analysed. In Błażkiewicz et al., (2014), participants performed 3 walking trials at a self-selected pace and step length, step duration, % stance phase, % loading response, % single support, % pre swing and % swing were analysed. Symmetry was calculated using RI, SI, GAsym and SA (Table 3.1).

Table 3.1 Indices for calculating gait asymmetry

Name	Equation
Ratio Index (RI)	$RI = \left(1 - \frac{X_R}{X_L}\right) \times 100\%$
Symmetry Index (SI)	$SI = \frac{\lvert X_L - X_R \rvert}{0.5 \times (X_L + X_R)} \times 100\%$
Gait Asymmetry (GAsym)	$GAsym = \ln\left(\frac{X_R}{X_L}\right) \times 100\%$
Symmetry Angle (SA)	$SA = \frac{45° - \tan^{-1}\left(\frac{X_R}{X_L}\right)}{90°} \times 100\%$

Results suggested that individual equations do not appear to provide any unique differences and the SI, RI, GAsym and SA were highly correlated. There was a high similarity between RI and SI and therefore the authors support existing recommendations that the SI should be used as the most sensitive assessment of gait symmetry on the basis of SPT parameters in healthy subjects.

3.5 Gait Kinematics

Gait kinematics are calculated using a biomechanical model which is normally comprised of a marker set, a method of calibrating a participant and a method of calculating the kinematics. Some models may also require anthropometric measurements to supplement the marker set and aid in calculation of joint centres. The following sections discuss each element of the biomechanical model in more detail.

The marker set used is the basis for tracking segments and subsequently, calculating kinematics. Therefore, placement of the markers must allow accurate representation of lower limb segments and markers must be placed in a manner such that an orthogonal axis system can be derived from their positions. Depending on the method of calibration, markers may also require placement which allows estimation of joint centre location from their position. Cappozzo et al., (1995) outline a number of prerequisites for optimal marker placement: the distance between markers should be sufficiently large to limit error propagation and they should not be co-linear as they are required to define a plane. Further, relative movement between markers and underlying bone should be minimal.

A number of commercially available marker sets exist, usually developed by motion capture companies. However, some labs may adapt marker sets to suit their specific clinical needs. One of the most widely used marker sets in the clinical environment is the Vicon Plug in Gait (PiG) model which has developed from the Helen Hayes (HH) model (Figure 3.4; Davis et al., 1991).

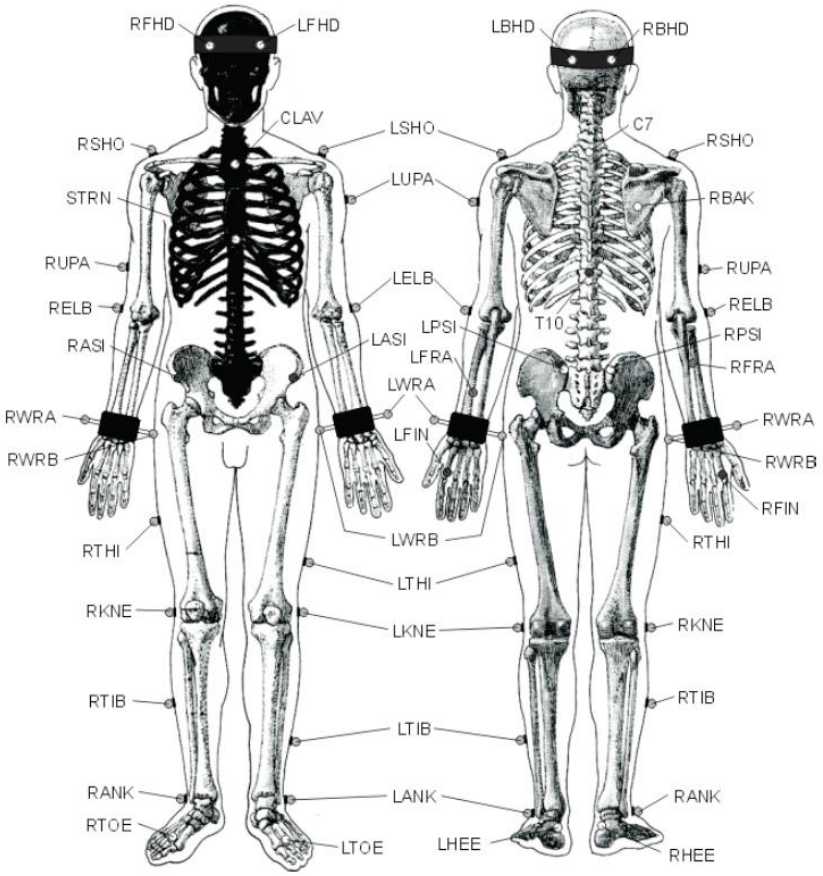


Figure 3.4 Standard full body marker set for Plug in Gait Protocol (Vicon Motion Systems, Oxford, UK)

An advantage of this model is that it has been well established within the clinical community, and as a result, is usually used for complex clinical cases (Schwartz and Rozumalski, 2005). Purpose written software for the extraction of clinically meaningful results has been developed and is relatively quick and straightforward to obtain in comparison to less well-established marker sets. However, there are a number of disadvantages associated with the PiG marker set. PiG requires markers placed directly on bony landmarks during dynamic trials and it has been suggested that markers placed on bony landmarks are subject to an unacceptable amount of movement during dynamic trials (Cappozzo et al., 1996). Further, segmental markers must be placed in very precise locations or errors may be introduced into the

kinematic output (Ferrari et al., 2008; McGinley et al., 2009). For example, the following extract is taken from the PiG marker placement manual and concerns the placement of the left thigh marker:

"Place the marker over the lower lateral 1/3 surface of the thigh, just below the swing of the hand, although the height is not critical. The antero-posterior placement of the marker is critical for correct alignment of the knee flexion axis. Try to keep the thigh marker off the belly of the muscle, but place the thigh marker at least two marker diameters proximal of the knee marker. Adjust the position of the marker so that it is aligned in the plane that contains the hip and knee joint centres and the knee flexion/extension axis" (Vicon Motion Systems, 2016).

Placement of the markers following these directions can be extremely difficult to do visually and if the thigh markers are not placed correctly then the definition of the flexion axis of the knee will be incorrect. This can often lead to crosstalk which occurs when the ARF from a segment is out of alignment with the axis about which rotations actually occur and therefore one kinematic output is mistaken for another (Piazza and Cavanagh, 2000). In this case, flexion/extension is often mistaken for abduction/adduction which will affect results. The Human Body Model (HBM; Motek Medical, Amsterdam) is another example of a commercially available marker set. HBM also employs individually placed skin surface markers (Figure 3.5), one of which is a marker placed over the greater trochanter.

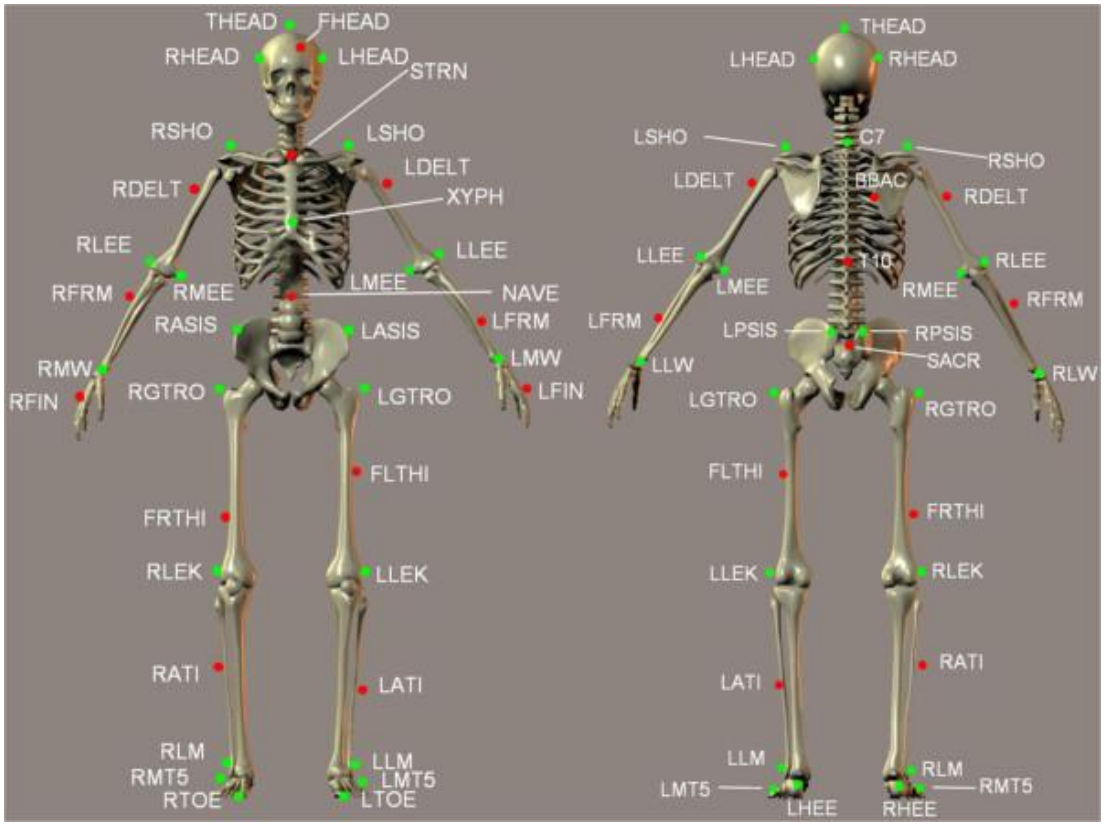


Figure 3.5 Human body model marker set (Motekforce Link, Netherlands)

However, evidence suggests palpation of the greater trochanter is prone to the greatest error (up to 18mm RMS) out of a number of lower limb anatomical landmarks (Della Croce et al., 2005). The mathematical modelling behind HBM is not currently available due to commercial sensitivities. However, it may be speculated that the greater trochanter is unlikely to be used in calculation of the HJC as the marker set also includes ASIS and PSIS markers which are far more suitable anatomical landmarks for HJC calculation. It is likely that the greater trochanter marker is used for calculation of the thigh segment ARF. Even so, as the location of the greater trochanter can be prone to error, it is unlikely to be a suitable marker for ARF calculation and could introduce inaccuracies into hip and knee kinematic outputs.

In conjunction with the two marker sets mentioned, the majority of commercially available marker sets employ individual skin surface markers. There is extensive evidence to suggest that single markers located on the skin are not the most accurate way to estimate segment position during dynamic trials as they introduce error due to relative movement between skin and underlying bone (Benoit et al., 2006; Leardini et al., 2005; Peters et al., 2010). This issue is often referred to as soft tissue artefact (STA). The most accurate method of measuring segment movement is to insert bone pins directly into the bones of the segments of interest and attach a triad of markers to the bone pin. This method allows measurement of actual bone movement and eliminates STA. A number of researchers have conducted studies using bone pins (Fuller et al., 1997; Karlsson and Lundberg, 1994; Reinschmidt et al., 1997); however, it is not an acceptable method for standard practice. Therefore, efforts must be taken to reduce interference from soft tissues by attaching markers as rigidly as possible in a non-invasive manner. STA is particularly hard to counteract when using non-invasive methods as displacement caused by skin has the same frequency content as that of the actual bone and therefore it is difficult to distinguish which is which (Cappozzo, 1991).

A number of authors have attempted to quantify the effects of STA on various kinematic outputs and marker positions. Sati et al., (1996) used x-ray and fluoroscopy to quantify skin movement at the knee. A number of stainless steel bearings were attached to the medial and lateral epicondyles and the lateral thigh. Slow, dynamic knee flexions were performed and marker movement varied from 2.5 mm root mean square (RMS) to 17 mm RMS with the largest deviations occurring when markers were placed on the joint line. Peters et al., (2010) conducted a review

of STA for skin surface markers and cluster based markers. It was suggested that markers over anatomical landmarks on the thigh exhibited significant STA (> 10mm) with the lateral epicondyle being particularly susceptible, exhibiting errors of up to 20 mm. However, cluster markers located elsewhere on the thigh seemed less prone to error (7-12 mm). Markers on the tibia were less susceptible to STA (3-15 mm displacement) than markers on the distal thigh and therefore it was concluded that the effects of STA are dependent on the segment under analysis.

Manal et al., (2000) investigated the effect of STA on kinematic estimates of knee internal and external rotation values. The effect of location, physical characteristics and attachment method were examined and results were compared to those obtained using individual markers. All rotational differences were relatively small (maximum of 2°); however, differences were smaller when marker arrays were placed more distally than proximally on the segment. The smallest errors were seen when using a constrained cluster array of markers on the distal part of the segment (Figure 3.6B).

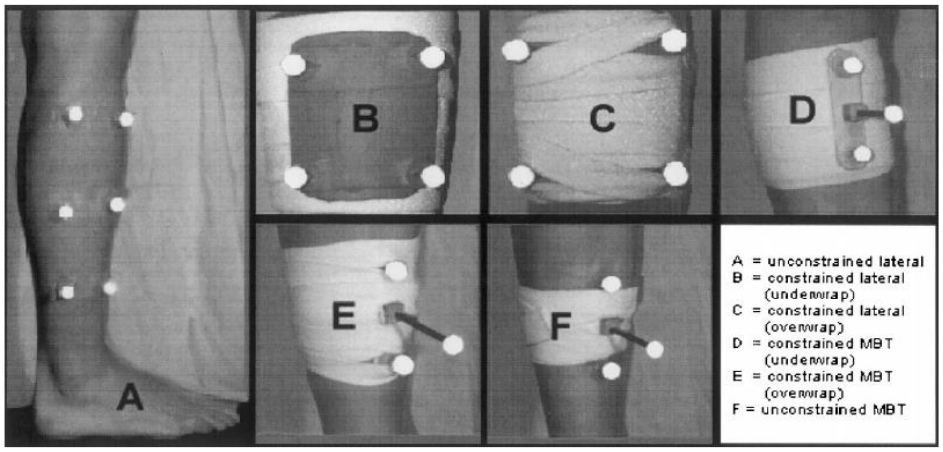


Figure 3.6 Variations in marker arrays investigated and the position of array on segment. Arrangement B resulted in the lowest error in rotational output (Manal et al., 2000)

Stagni et al., (2005) investigated STA on the thigh and shank using traditional motion capture compared to 3D kinematics reconstructed from fluoroscopic images.

A large number of markers were attached to the thigh and shank to allow options regarding which would be used for calculation of kinematics (Figure 3.7).

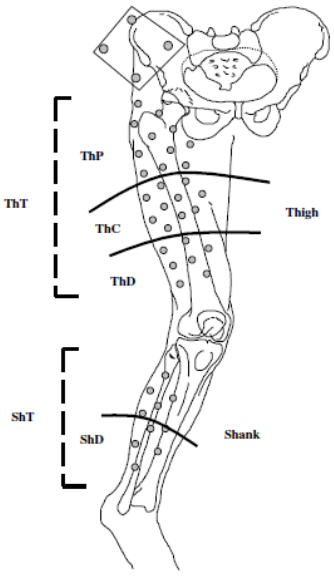


Figure 3.7 Skin surface marker distribution. Markers are divided into proximal, central or distal clusters on the thigh and distal or total clusters on the shank (Stagni et al., 2005)

All skin surface markers exhibited considerable movement; however, shank markers were subject to less STA than thigh markers. Maximum deviation of shank markers was 20.6 mm and maximum deviation of thigh markers was 31.1 mm. When choosing different clusters of markers for kinematic calculation, large differences were seen depending on which cluster was selected. It was suggested that abduction/adduction and internal/external rotation are more affected by STA and are critically dependent on cluster selection. The maximum RMS error for flexion/extension was 23.4%; for abduction/adduction it was 191.8% and for internal/external rotation it was 116.6%. It was concluded that joint kinematics are strongly affected by the choice of cluster markers as a result of STA error propagation. It is suggested that thigh markers should be placed more distally to reduce STA.

Leardini et al., (2005) conducted a review of marker displacement and kinematic measurement errors due to STA. A summary of the findings are detailed in Table 3.2.

Table 3.2 Summary of papers investigating STA from Leardini et al., (2005)

Authors	Participants	Gold standard	Measurement techniques for comparison	Results
Lafortune and Lake, (1991)	1	X-ray Fluoroscopy	Marker on proximal tibia	21mm distal and 23mm posterior displacement of marker from bone surface
Karlsson and Lundberg, (1994)	2	Bone screws anchoring marker triad to femur and tibia	3 skin markers on the thigh and shank	20° knee rotation when measured with bone mounted markers and 50° when measured with skin mounted
Reinschmidt et al., (1997)	3	Intra-cortical Hoffmann pins anchoring marker triad to femur, tibia and calcaneus	6 skin makers on thigh, shank and shoe	FL/EX showed good agreement. AB/AD and IN/EX showed errors as high in magnitude as the real joint motion
Fuller et al., (1997)	1	6 markers inserted into the tibia and femur	20 markers on the thigh and shank	Skin maker displacement of up to 20 mm
Westblad et al., (2000)	3	Markers anchored to tibia, fibula, talus and calcaneus to measure ankle complex motion	3 markers on shank, heel and forefoot	Mean maximal differences in joint rotations were less than 5°

These studies highlight the significant issues which arise when using skin surface markers in an attempt to measure kinematics. Evidence suggests that STA can be reduced by using rigid plates of markers and refraining from placing markers too proximally on the thigh.

3.5.1 Participant Calibration

Participant calibration is an extremely important aspect in the measurement of human kinematics. While the accuracy of kinematic measurements is dependent on a number of factors, participant calibration plays a key role in the output of meaningful kinematic data. The purpose of participant calibration is to allow estimation of joint centre locations and determine the ARFs for each participant whose movement is to be recorded. If a cluster based marker set is being used, participant calibration also involves calculating the position of anatomical landmarks relative to the cluster which allows reconstruction of anatomical landmarks from the position of the cluster during dynamic trials. This will subsequently allow the creation of dynamic ARFs and calculation of kinematics.

The commercially available marker sets discussed in previous sections allow participant calibration using the marker set combined with a number of anthropometric measurements. This is due to the marker set implementing markers placed on key anatomical landmarks. With a cluster based marker set there are no markers on anatomical landmarks so alternative calibration methods must be devised. The following text describes a number of participant calibration methods which can be implemented when using a cluster based model.

There are 3 main methods of participant calibration: static, functional and pointer. Static calibration requires the placement of skin surface markers on the anatomical

landmarks of interest while at least one frame is recorded. One advantage of this method is simultaneous estimation of joint centres and calculation of anatomical landmark positions relative to the cluster in one frame. This allows all calculations to be carried out quickly and simultaneously. However, evidence suggests placement of markers on bony landmarks should be avoided where possible as this can lead to high levels of error due to STA (Cappozzo et al., 1996). Even though the calibration is completed with the participant in a static pose, the possibility still exists for marker movement between the placement of all the markers and the participant adopting the calibration pose. Further, it is a time consuming process to apply a number of skin surface markers for the purpose of collecting only a few frames of data and skin surface markers require minimal clothing to be worn which is not ideal for routine clinical use.

Functional calibration requires the participant to perform a ROM at each joint. However, it was not deemed suitable for this study as a number of participants are likely to have reduced ROM at one or more joints and may not be capable of performing the movements required for a suitable functional calibration. Advantages and disadvantages of functional calibration are discussed in more detail in the following section.

Pointer calibration is similar to static calibration as it requires the location of anatomical landmarks to estimate joint centres. However, the method by which the position of anatomical landmarks is determined is slightly different. Instead of placing skin surface markers on the anatomical landmarks of interest, a pointer which has a tip of known position is used. The user places the tip of the pointer at the anatomical landmark of interest and the position of the anatomical landmark, along

with its location relative to the corresponding cluster, is stored. One advantage of this method, particularly for clinical use, is that it does not require the participant to wear minimal or skin tight clothing, as long as appropriate clothing is worn to allow temporary palpation of the anatomical landmarks of interest. Further, it negates the use of skin surface markers which may lead to reduced errors during the calibration stage.

Joint centre location is achieved by functional methods or by identifying the position of anatomical landmarks surrounding the joint and, using well defined regression equations (Bell et al., 1989; Harrington et al., 2007), estimating the position of the joint centre relative to these landmarks. Accurate estimation of joint centre locations is essential as they form the basis of the ARF. As detailed in chapter 2, when functional calibration is not employed, estimation of the KJC and AJC is achieved by taking the midpoint of the relative LE and ME for the knee and LM and MM for the ankle.

The position of the HJC can be estimated using functional ROM based methods or using regression equations which relate the HJC to pelvic anatomical landmarks. The functional method relies on tracking the movement of one or more markers on the thigh as the participant performs a ROM task at the hip. Sphere fitting algorithms are then used to determine the centre of the sphere created by movement of the thigh markers. The centre of the sphere is assumed to be the HJC. However, functional methods often require the participant to perform large and sometimes lengthy ROM tasks which may not be appropriate for patients with pain or limited ROM (Bell et al., 1990). Further, the methods used to locate the HJC when using the functional approach can have an effect on the accuracy of the estimation of its

position (Piazza et al., 2001). Therefore, a number of considerations should be taken into account when choosing which method should be employed for estimation of the HJC.

Bell et al., (1990) investigated the accuracy of functional HJC estimation compared to the accuracy of a regression equation developed by Andriacchi and colleagues. The gold standard for HJC estimation was a pair of orthogonal radiographs. The functional method estimated the HJC to be an average of 3.79 cm from the true HJC whereas the regression equation estimated the HJC to be an average of 3.61 cm from the true HJC. It was concluded that neither method was particularly accurate for estimation of HJC location and the errors could introduce significant artefacts into movement data. Further, the functional method required participants to sequentially flex, extend and abduct the leg, passing through the anatomical position each time whilst keeping the long axis of the foot orientated in the anterior-posterior direction. This ROM task may prove difficult for patients who have reduced stability, limited ROM or pain in the hip, and therefore is unlikely to be the most effective method for estimating HJC location in the clinical setting.

Piazza et al., (2001) aimed to determine how the implementation of functional HJC estimation affects accuracy. A number of variations on the standard ROM task were tested, including: increasing the number of hip motion observations collected, use of both planar and 3D hip motions and limitation of hip motion to a range that might be expected in patients with hip pathology. A mechanical linkage capable of mimicking hip movement was used to test a number of implementations. When ROM was reduced from 30° to 15°, there were significant increases in the magnitude of HJC estimation error. However, there was no significant difference when ROM was

maintained at 30° but the type of motion was altered. The largest error in HJC location estimation was seen when ROM was limited to 15° and motion was limited to circumduction. Further, trials performed with increased duration yielded almost twice the number of data points for sphere fitting algorithms but estimation of HJC location was almost identical. Use of the standard functional method resulted in a mean HJC location error of 4.4 mm. However, it is highly likely that errors reported in this study are not representative of HJC estimation errors in human subjects, as a mechanical linkage was used which would exhibit no STA or muscular contraction, therefore greatly reducing any STA errors which would be present in human subjects. What can be derived from the study is that a reduced ROM is likely to have a significant effect on the estimation of HJC location, suggesting again, that this method may not be the most appropriate option for patients who may not be able to fully complete the ROM task.

Piazza et al., (2004) also aimed to investigate the effects of reduced ROM and using functional tasks such as walking and stair climbing to estimate HJC location. The average error from a reduced ROM estimation was 26 mm, whereas the average error from functional tasks was 70 mm. However, the gold standard in this study was a full ROM functional estimation of HJC location and evidence has suggested this is also prone to error. Therefore, it is possible that the true error measured from limited ROM and functional tasks is even larger than that reported, meaning these are unlikely to be appropriate methods for implementing the functional approach.

Kainz et al., (2015) carried out an extensive review of methods to estimate the hip joint centre. It was noted that some functional methods resulted in slightly higher repeatability and were more reproducible than regression methods based on

anatomical landmarks. It was also stated that increasing the number of samples and ROM may result in a more precise estimation of HJC location; however in some cases reduced ROM may result in errors of less than 10 mm. Further, it was reported that single plane motions increase errors in HJC estimation when using functional methods. The average error reported from the most effective functional method (geometric sphere fit) was 11 to 21 mm and to reach this level of precision participants should be able to achieve at least 60° flexion extension and 30 - 40° ab/adduction.

As evidence suggests the functional method requires a large ROM which many patients may not be able to achieve, a number of well-defined regression equations have been developed which allow estimation of HJC location from pelvic anatomical landmarks (Bell et al., 1989; Davis et al., 1991; Harrington et al., 2007). Harrington et al., (2007) assessed 3 popular regression equations for estimation of HJC location in an attempt to quantify errors involved in their application to healthy adults and children and to children with cerebral palsy (CP). The gold standard for HJC estimation was MRI scans. The regression equations investigated were (Davis et al., 1991), Bell et al., (1989) and software recommendations for OrthoTrack motion capture software. Results suggested that the Bell and OrthoTrak methods gave HJC locations closest to those of MRI scans with maximum errors of 21mm and 18mm, compared to 26 mm using the Davis method. It was suggested that pelvic depth (PD) was the most suitable predictive variable for HJC location in the anterior-posterior and medio-lateral directions, whereas pelvic width (PW) was most suitable for estimation in the superior-inferior direction. It was also suggested that prediction

errors of this magnitude are likely only to have a small effect on resulting joint angles.

3.5.2 Anthropometric Measurements

A number of commercial marker sets, including PIG and HBM, require anthropometric measurements to augment marker placement. These measurements may include height, leg length, distance between the right and left anterior superior iliac spines (inter ASIS distance), knee width and ankle width. Knee and ankle width are required for the calculation of the KJC and AJC, respectively, due to the absence of medial markers on the knee and ankle. The PIG and HBM marker sets are used for both static and dynamic trials. If medial markers were used on the knee and ankle during dynamic trials, there is a high possibility that they would be knocked off due to the proximity of these landmarks. However, anthropometric measurements may not be the best method to determine joint centres as the accuracy of the measurement depends on the measurement device and the user, which could vary between participants or sites, leading to inconsistencies in data (Klipstein-Grobusch et al., 1997).

3.5.3 Kinematic Calculation

The ARF of each segment is the basis for kinematic calculation. The method of calculating kinematics may vary between biomechanical models, but in order to calculate joint angles, the axes from the ARF of the segment proximal and distal to the joint are required. By using either classic mechanics or modified classic mechanics methods (Cole et al., 1993; Grood and Suntay, 1983), the angles between segments can be obtained. Kinematic calculation is discussed in more detail in section 3.11.

3.6 Gait Protocols

A gait protocol is defined as a biomechanical model and the associated procedures for data collection, reduction and analysis of results (Ferrari *et al.* 2008). The majority of protocols currently used have been developed based on the Newington Model. This model was developed over 20 years at the Newington Children's Hospital and is the minimal configuration needed for 3D, bilateral analysis of gait (Figure 3.8; Davis et al., 1991). Further, it was developed at a time when using the lowest number of markers possible was required due to the limitations of the technology for motion capture 25 years ago.

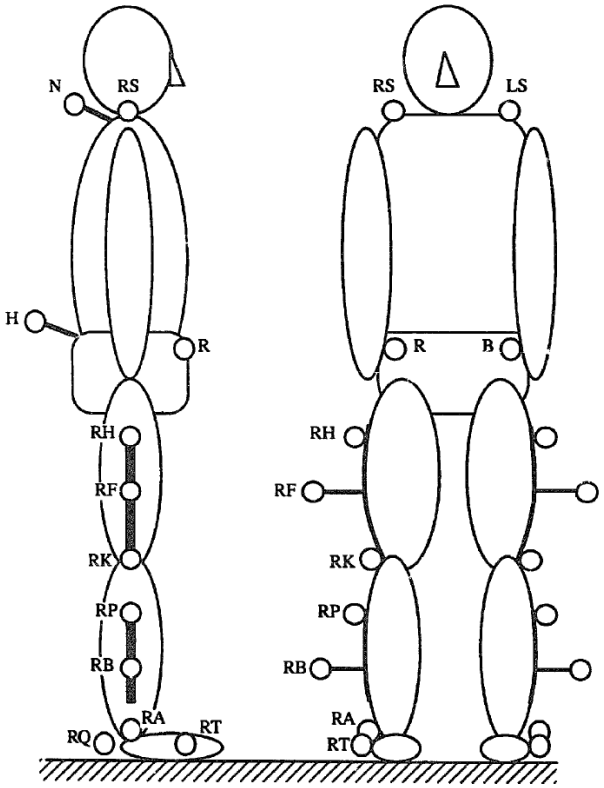


Figure 3.8 Newington marker set (Davis et al., 1991)

Specific clinical interests may require labs to develop their own unique protocols. However, this is uncommon for routine clinical practice as a detailed understanding of biomechanical mathematics is required as well as a thorough technical

understanding of motion capture hardware and software, hence the Newington model has continued to be the most widely used protocol for clinical and research applications.

However, there are a number of factors which may vary between GA protocols. Different marker sets may be employed, which will result in different definitions of anatomical and technical reference frames. Further, joint centre estimation and joint angle calculation and definition may also vary. It would therefore be unsurprising to observe differences in the kinematic output from different GA protocols. Despite this fact, GA data is often compared between sites and institutions which may implement different protocols. There has been limited investigation into the differences between kinematic outputs from different GA protocols. A limited number of authors have compared protocols to determine if any discrepancies in data exist.

Stief et al., (2013) aimed to investigate the reliability and accuracy of two GA protocols. A custom designed protocol (MA) was compared with the current clinical gold standard; PiG, which is a commercial version of the Newington or Helen Hayes (HH) model. A single comprehensive marker set was established (Figure 3.9a) and average data from 25 subjects performing 5 walking trials each was captured. Results showed average differences of up to 8° in knee ab/adduction and up to 5° in knee flexion/extension. Both differences were significant ($P<0.05$). It was suggested that differences were caused by anatomical landmark palpation errors and different placement of markers between the two protocols.

Ferrari et al., (2008) aimed to assess the inter protocol variability of five GA protocols: PiG, the protocol developed at the 'Servizio di Analisi della Funzione

Locomotoria (SAFLo), Calibration Anatomical System Technique (CAST), the protocol 'Laboratorio per l'Analisi del Movimento nel Bambino' (LAMB) and the basis of the Total 3D Gait (T3D) software. Again, a single comprehensive marker set was devised which incorporated all marker sets, allowing simultaneous capture for each walking trial (Figure 3.9b). Two able bodied subjects and one subject with a knee prosthesis which prevented ab/adduction were tested. Overall, gait variables were comparable between models (Figure 3.10) despite differences in the marker sets. The kinematic output showed larger discrepancies than kinetics, with out-of-sagittal plane rotations at the knee and ankle showing significant differences. The largest difference was observed in knee ab/adduction. It was hypothesised that this was caused by kinematic crosstalk between the flexion/extension and ab/adduction axes of the femur and tibia which is likely to be caused by misplacement of the lateral thigh marker causing the knee flexion axis to be defined incorrectly.

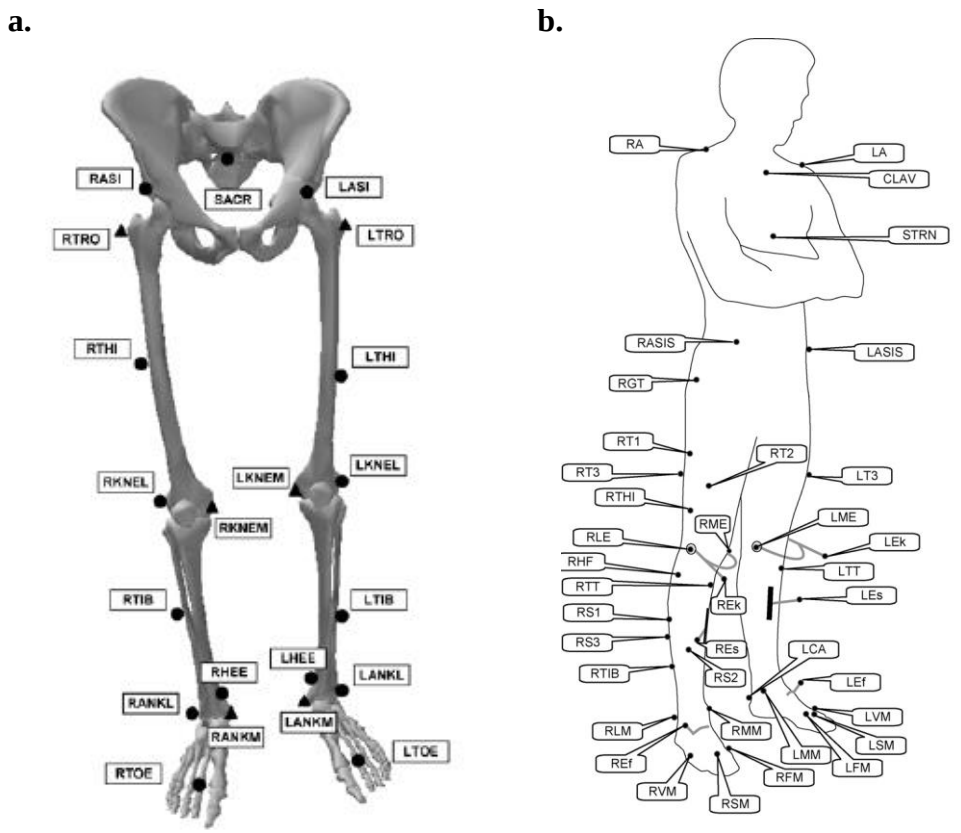


Figure 3.9 Comprehensive marker sets for comparison studies **a.** Marker set for MA and PiG protocols. Markers indicated by circles are part of PiG model and those indicated by triangles are part of the MA protocol (Stief et al., 2013) **b.** Marker set for 5 GA protocols (Ferrari et al., 2008)

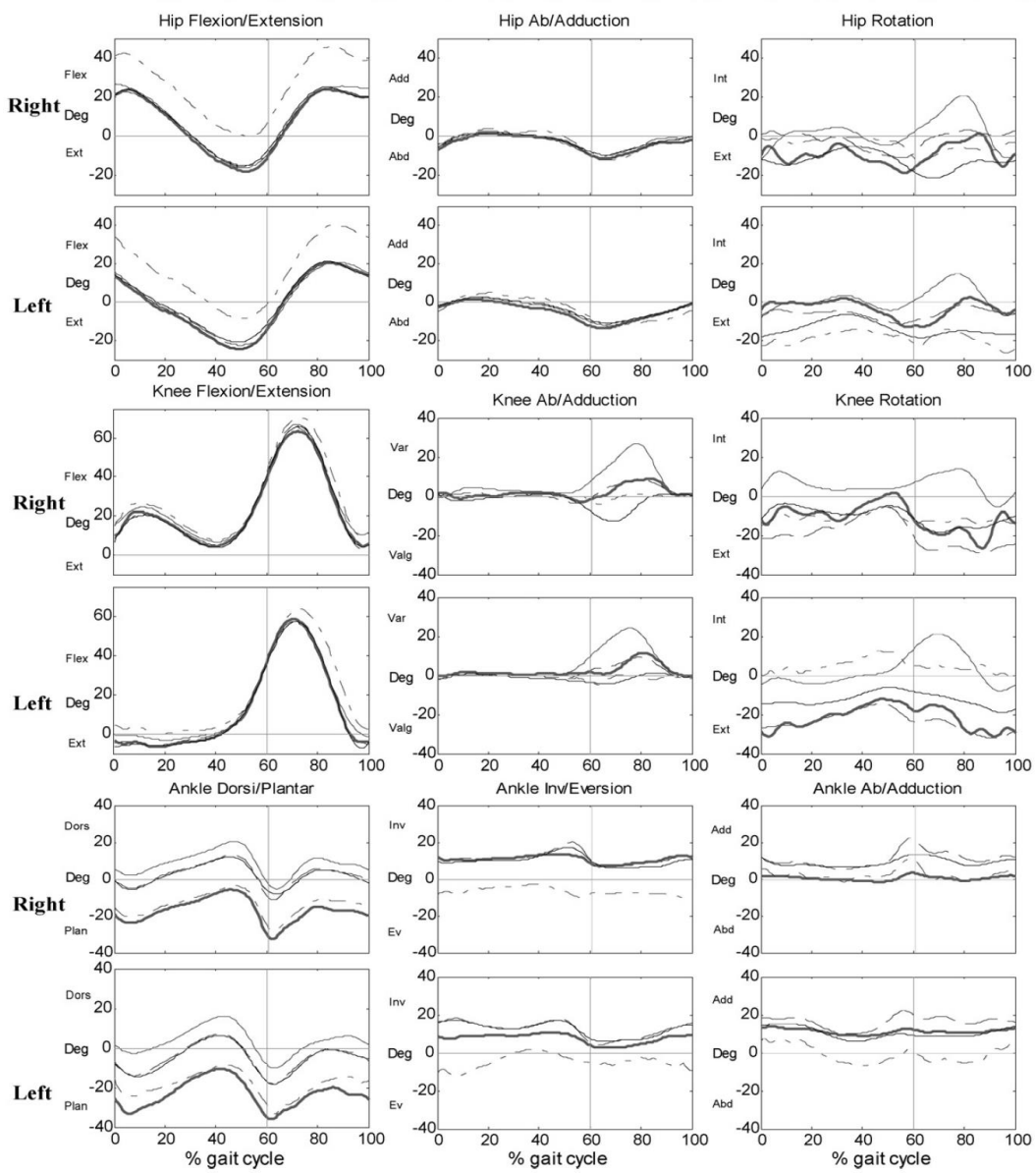


Figure 3.10 Kinematic variables as calculated by the five protocols and relative to one subject. T3D (dash), PiG (dot lines), CAST (black solid), SAFLo (dash-dot), LAMB (grey thick-solid) (Ferrari et al., 2008)

For both these studies, it was hypothesised that differences in the kinematic output were caused by different marker locations. As described in chapter 2, the location of skin surface markers will ultimately define the ARF and as the ARFs are used to calculate kinematics, it is unsurprising that discrepancies are seen between marker sets.

3.7 Development of Bespoke Cluster Based Marker Set

Based on the evidence presented in the previous section, a bespoke cluster based marker set using rigid plastic plates was presumed to be the best option for a routine clinical marker based motion tracking tool. Further, a pointer based calibration method was devised as this negates the need for skin surface markers or lengthy and complex ROM tasks. The following section details development of the marker set, development of the pointer calibration method, calculation of kinematics and a comparison of the bespoke Strathclyde Cluster Model (SCM) to the current clinical gold standard.

Figure 3.11 shows the complete cluster marker set. Cluster plates were designed using computer aided design software (Creo parametric 3.0, PTC Needham USA) and 3D printed using an Ultimaker 2 extended 3D printer (Ultimaker B.V). Each plate was designed to be as small as possible in order to maximise comfort for the wearer while still allowing space for 4 well placed markers. A small amount of curvature was introduced into the thigh and shank plates to increase the congruency of the plate with the segment and prevent excess movement. Figure 3.11 shows the approximate location of the plates on each segment.

Figure 3.11 Placement of plates on each segment. **a.** Thigh and shank plates are attached using Velcro straps **b.** Pelvis plate is placed on the back, below the level of the PSISs **c.** Approximate location of thigh, shank and foot plates

As the calibration stage calculates the position of anatomical landmarks relative to clusters, the plates do not need to be placed accurately on the segment. However, there are a few stipulations which should be followed. The pelvis plate must be positioned inferior to the PSIS anatomical landmarks. As these are the most superior anatomical landmarks of the pelvis, a plate placed above these anatomical landmarks will technically be measuring trunk movement, not pelvis movement. Further, as evidence suggests markers placed more proximally on the thigh may exhibit higher levels of STA, it is advised that thigh plates are placed more distally. It is also advised that shank plates are placed distally to avoid interference from the gastrocnemius muscle during dynamic trials. Further, all plates must be placed on the correct segments and the correct way up.

In order to estimate joint centre locations and determine segment ARFs, it is necessary to ensure all clusters are identified and labelled correctly. Initially, Vicon

Nexus was used to stream marker trajectories into D-Flow, which didn't allow automatic identification of bespoke clusters. Therefore, a custom cluster recognition and labelling algorithm was written in Lua code.

Each cluster was designed with a unique marker arrangement, thus allowing the distances between markers to be used for cluster identification. Prior to tracking, the x, y and z position of each marker on each cluster was stored which allowed calculation of the 6 inter-marker distances. These were then compared to inter marker distances during dynamic trials to allow identification of the correct cluster.

During dynamic trials, the distance between each marker and all other markers in the capture volume was calculated. For each marker, the 4 closest markers were identified and for each set of 4 markers, the 6 inter-marker distances were calculated. These distances were then compared to the stored data for each cluster to determine which 6 distances best match those which have been previously stored. A tolerance of 2 mm between each measured and stored distance was applied which allowed robust recognition without confusion between clusters. However, there was a high computational demand required to run this code, and as a result the performance of the D-Flow software suffered. Therefore, Vicon Tracker was implemented as this allowed pre-identified clusters to be streamed into D-Flow. For each cluster, an 'object' was created in Tracker and a Vicon Skeleton Template was assigned to the object which allowed the cluster to be automatically tracked. Following creation of the 'object', clusters would continue to be tracked even if they were removed from and returned to the capture volume. Further, creation of each 'object' only had to be done once, and providing the same clusters were used, these were recognised by Tracker and streamed into D-Flow during every session.

Once clusters were located, individual markers were labelled. All markers were labelled as described in Figure 3.12 which allowed all TRFs to be created in the same way, as described in chapter 2.

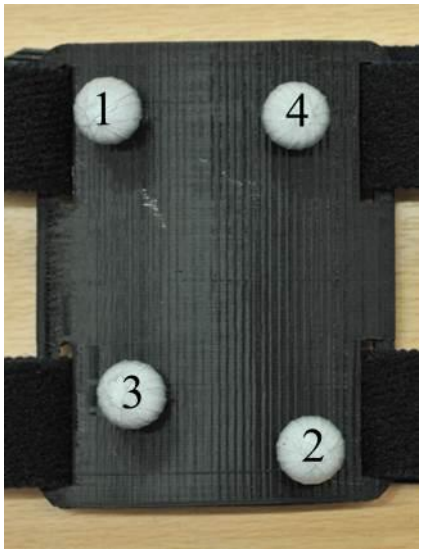
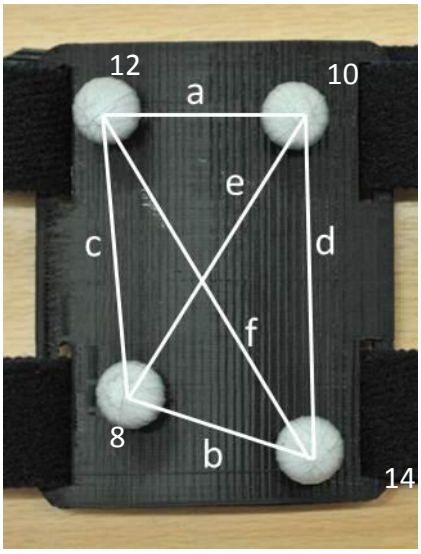


Figure 3.12 Marker labels applied to all clusters

In order to determine which marker was which, the distances between the markers were calculated and sorted into ascending order, returning a table of each inter marker distance and a number index which was assigned to each marker by D-Flow. An example is shown in Figure 3.13.



Distance	Length (m)	Marker Indices	
a	0.0452	12	10
b	0.0536	8	14
c	0.0622	12	8
d	0.0675	10	14
e	0.0724	10	8
f	0.0758	12	14

Figure 3.13 Example of marker labelling algorithm. Distances were labelled in ascending order from a to f. Marker indices were assigned by D-Flow in the order in which markers were recognised. Arbitrary indices have been chosen for this example. Distances and indices were constructed into a table in ascending order of distance length

Clusters were designed so that marker 1 was always at one end of the shortest distance and it was therefore known that one of the indices from distance *a* must be marker 1. In the example shown in Figure 3.13, marker 1 was also at the end of distances *c* and *f*. Therefore marker 1 must be the index which appeared next to distances *a*, *c* and *f*, which in this example, is 12. Once marker 1 was located, it could be determined that marker 4 must be the other index next to distance *a*, as this was the marker at the other end of the shortest distance. To find the remaining 2 markers the other indices from distance *c* and *f* were used. This was repeated for each cluster and allowed all cluster markers to be labelled in a consistent manner. The code for marker labelling is presented in appendix 1.

Markers were positioned such that each inter marker distance was at least 2mm greater than the previous inter marker distance, thus ensuring that the order of inter marker distances did not change with small fluctuations in distance caused by noise in the system.

3.8 Development of Pointer Based Calibration

Initially, anatomical landmarks were located using skin surface markers during participant calibration. However, this method was rejected due to evidence suggesting the high level of error which could be introduced (Cappozzo et al., 1996; Leardini et al., 2005). Further, skin surface markers require participants to wear tight or minimal clothing, which is not ideal in the clinical environment. Therefore, skin surface markers were rejected in favour of a pointer based calibration method.

The pointer in this study was developed as a cluster of 4 markers (Figure 3.14). This allowed the position and orientation of the pointer to be known at all times. It also allowed tracking and labelling in the same way as was done for the body segment clusters.

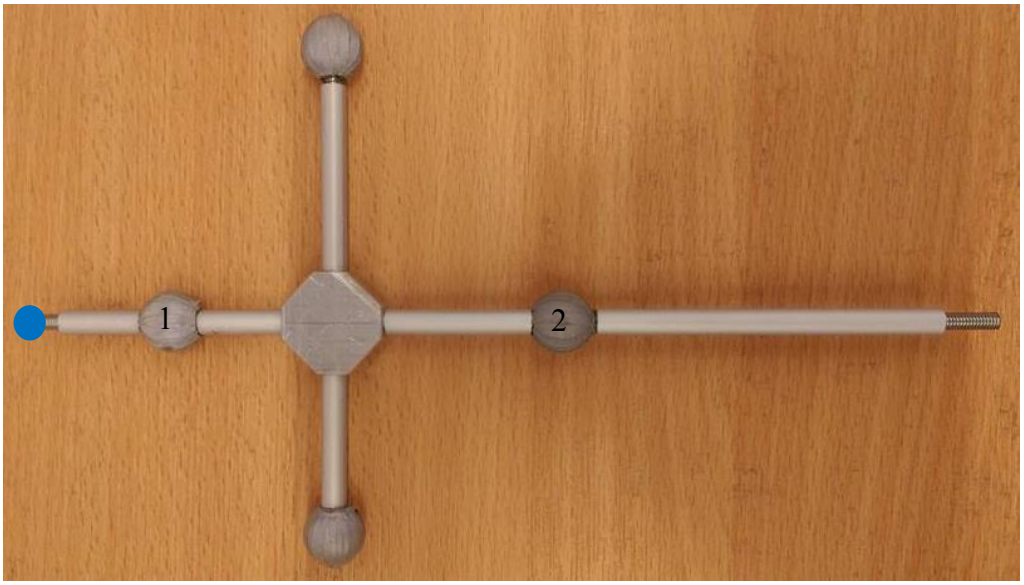


Figure 3.14 Pointer used to locate anatomical landmarks. Blue dot indicates tip which is placed on the anatomical landmark of interest

The tip of the pointer was at a known distance from markers 1 and 2 and was reconstructed using a simple vector extension algorithm (Equation 3.1; appendix 1) where M is the distance between marker 2 and the pointer tip, L is the distance

between marker 1 and marker 2, V_1 is the vector between markers 1 and 2 and V_2 is the vector between marker 2 and the pointer tip.

$$\begin{bmatrix} x \\ y \\ z \end{bmatrix}_{V_2} = \begin{bmatrix} x \\ y \\ z \end{bmatrix}_{V_1} \times \left({}^{M}/_{L}\right)$$

Equation 3.1

In order to determine the accuracy of anatomical landmark location using a pointer, a test was carried out to determine the effect of pointer orientation on resulting anatomical landmark position and subsequent calculation of kinematics.

3.8.1 Methods

This test was carried out using a 12 camera Vicon motion capture system (Vicon MX, Oxford Metrics Ltd., UK). The system consisted of 6 16 megapixel cameras and 4 4 megapixel cameras in fixed positions. All cameras sampled at 100Hz. Prior to capturing data, the system was calibrated using the methods described in chapter 2.

The highest residual was 1.6 mm and the mean residual was 0.99 mm. A mark was made on a box which was kept in a fixed position in the capture volume. The tip of the wand was placed on the mark and the x, y and z positions of the mark and orientation of the wand were stored at various wand positions. The wand was placed at four varying degrees of pitch, roll and yaw. Pitch was defined as rotation about the global X axis, roll was defined as rotation about the global Z axis and yaw was defined as rotation about the global Y axis (Figure 3.15 and Figure 3.16). Table 3.3 details the orientation of the wand at each point that the mark position was stored. The wand was also placed at 4 random combinations of pitch, roll and yaw and the mark position stored.

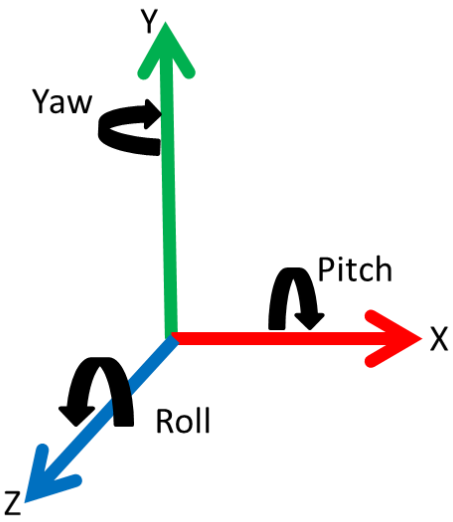


Figure 3.15 Pitch, roll and yaw as defined in the GRF

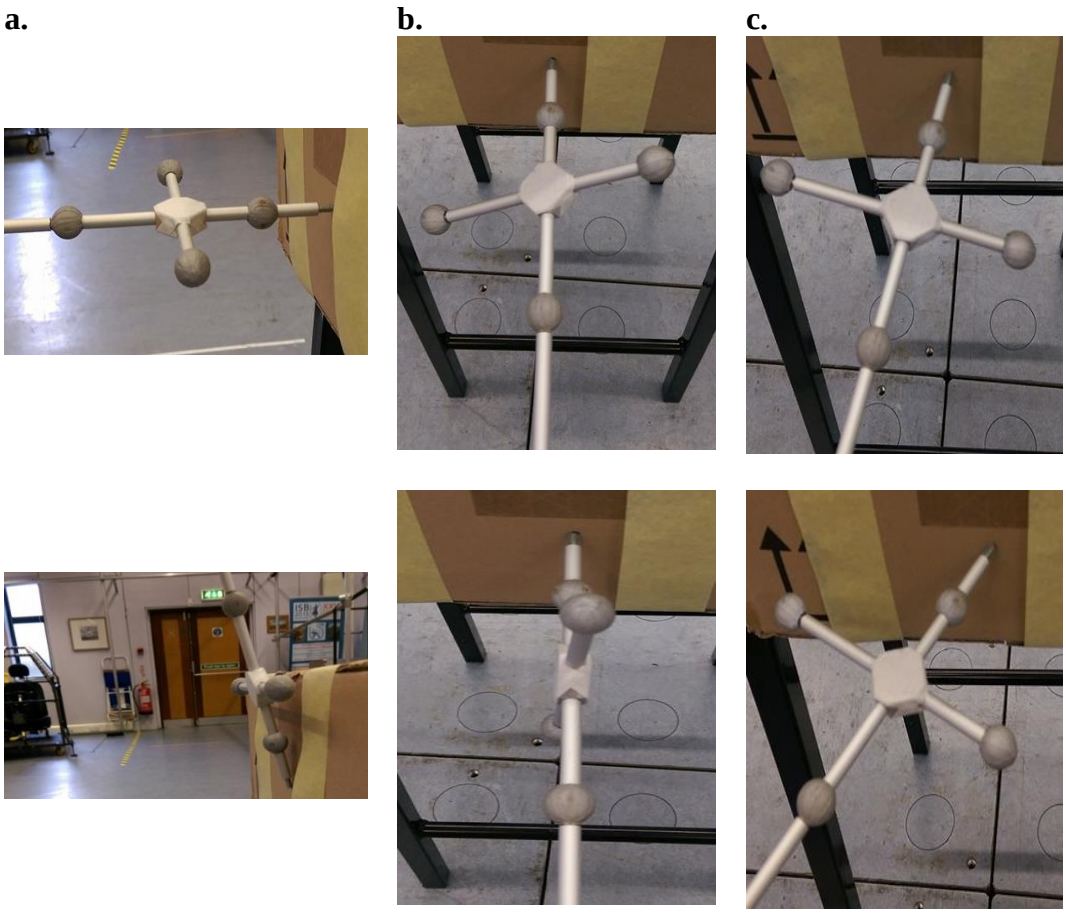


Figure 3.16 Example of different want positions **a.** Pitch **b.** Roll **c.** Yaw

Table 3.3 Wand positions at varying degrees of pitch, roll and yaw

Pitch	Degrees	Roll	Degrees	Yaw	Degrees
Position 1	17.0	Position 1	15.4	Position 1	14.6
Position 2	47.7	Position 2	45.4	Position 2	37.6
Position 3	61.1	Position 3	60.2	Position 3	48.3
Position 4	70.4	Position 4	71.7	Position 4	58.0

The difference in the x, y and z position of the mark and the Euclidean distance between each location was calculated between each position for pitch, roll and yaw. The mean absolute error was then calculated for each orientation. To determine the effect that any error might have on kinematic output, one walking trial from a healthy individual was processed 5 times, once with the lateral epicondyle marker in the correctly palpated position and 4 subsequent times moving the lateral epicondyle marker in the proximal, distal, lateral and medial directions by the magnitude of the highest error. Knee flexion/extension, ab/adduction and internal/external rotation outputs were compared to determine if there was any significant difference in kinematic output.

3.8.2 Results

Table 3.4 through Table 3.6 show the difference in marker position for varying degrees of pitch, roll and yaw, respectively.

Table 3.4 Difference in marker position for varying degrees of pitch. Diff – difference, ED – Euclidean difference

	X diff (mm)	Y diff (mm)	Z diff (mm)	**ED (mm)**
Position 1 → Position 2	3.17	-3.47	-2.92	**5.54**
Position 2 → Position 3	-0.54	-1.69	1.65	**2.42**
Position 3 → Position 4	-0.52	-0.74	0.51	**1.04**
Mean Absolute Error (mm)	1.69			

The largest Euclidean difference occurred when moving from position 1 to 2 with a total deviation of 5.5 mm. The smallest difference occurred when moving from

position 3 to 4. The mean absolute error for the difference in one direction was 1.69 mm.

Table 3.5 Difference in marker position for varying degrees of roll. Diff – difference, ED – Euclidean difference

	X diff (mm)	Y diff (mm)	Z diff (mm)	**ED (mm)**
Position 1 → Position 2	-0.59	0.48	1.03	**1.29**
Position 2 → Position 3	-0.25	-0.56	-1.07	**1.24**
Position 3 → Position 4	-0.41	-1.78	0.46	**1.88**
Mean Absolute Error (mm)	0.74			

The largest Euclidean difference occurred when moving from position 3 to 4 with a total deviation of 1.8 mm. The smallest deviation occurred when moving from position 2 to 3. The mean absolute error for the difference in one direction was 0.74 mm.

Table 3.6 Difference in marker position for varying degrees of yaw. Diff – difference, ED – Euclidean difference

	X diff (mm)	Y diff (mm)	Z diff (mm)	**ED (mm)**
Position 1 → Position 2	-0.52	0.17	0.63	**0.84**
Position 2 → Position 3	-0.09	-0.3	-0.89	**0.94**
Position 3 → Position 4	-1.08	-0.12	-0.18	**1.1**
Mean Absolute Error (mm)	0.44			

The largest Euclidean difference occurred when moving from position 3 to 4 with a total deviation of 1.1 mm. The smallest difference occurred when moving from position 1 to 2. The mean absolute error for the difference in one direction was 0.44 mm.

Table 3.7 shows the difference in marker position for four random combinations of pitch, roll and yaw.

Table 3.7 Difference in position for varying degrees of pitch, roll and yaw. Diff – difference, ED – Euclidean difference

	X diff (mm)	Y diff (mm)	Z diff (mm)	**ED (mm)**
Position 1 → Position 2	-5.42	1.67	0.56	**5.71**
Position 2 → Position 3	4.49	1.3	-0.9	**4.76**
Position 3 → Position 4	-3.81	-0.25	0.92	**3.93**
Mean Absolute Error (mm)	2.15			

The largest Euclidean difference was 5.7 mm and the mean absolute error for the difference in one direction was 2.15 mm.

The maximum deviation in one direction over all movements was 5.4 mm which occurred in the X direction when moving from random orientation 1 to random orientation 2. Figure 3.17 shows the kinematic output with the marker in the correctly palpated (benchmark) position and the effect of moving a marker 5.4 mm in the anterior, posterior, distal and proximal directions.

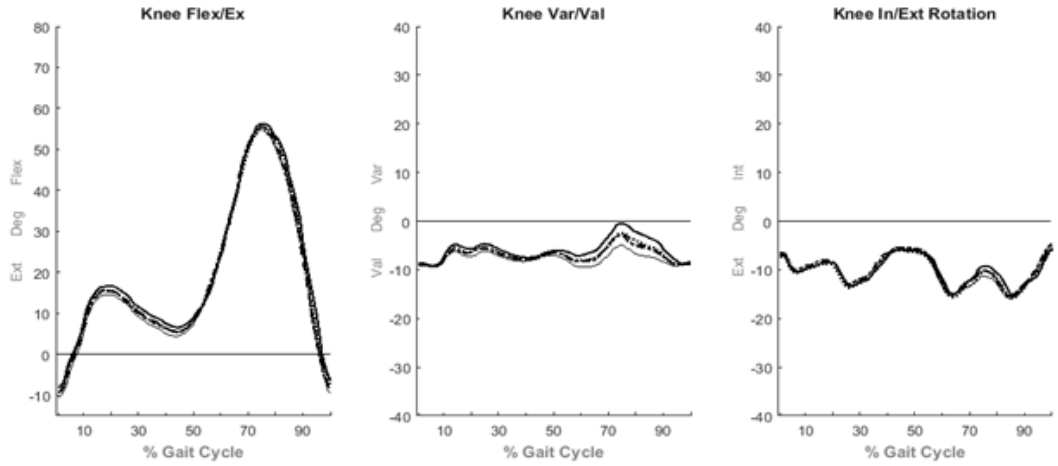


Figure 3.17 Effect of moving a marker 5.4 mm from the benchmark position (dashed) in the proximal (dash dot), distal (dots), anterior (thick solid) and posterior (thin solid) directions.

One way analysis of variance (ANOVA, $\alpha = 0.05$) was carried out to determine if there was a significant difference between the 5 scenarios for each kinematic output (Table 3.8). There was no significant difference for flexion/extension (P=0.953) or internal/external rotation (P=0.525) however, there was a significant difference for ab/adduction (P<0.001). To determine which deviations were significantly different,

a paired-sample t-test ($\alpha = 0.05$) assuming unequal variance was carried out on each deviation in comparison to the benchmark position for ab/adduction (Table 3.9). There was a significant difference for all deviations except proximal (P=0.939).

Table 3.8 ANOVA results for each kinematic output *significant difference ($\alpha = 0.05$)

Kinematic Output	P Value
Flex/Extension	0.953
Ab/adduction	**<0.001***
Int/Ext Rotation	0.525

Table 3.9 T-Test results for each deviation from the benchmark position *significant difference ($\alpha = 0.05$)

Deviation	P Value
Anterior	**<0.001***
Posterior	**<0.001***
Distal	**<0.001***
Proximal	0.939

3.8.3 Discussion

The aim of this test was to determine if pointer orientation had an effect on marker position and subsequent kinematic calculation. Results suggest that changing the pointer orientation does have a small effect on marker position as the largest deviation in one direction was 5.4 mm, occurring in the X direction. This deviation occurred between 2 random pointer orientations. The maximum deviation in one direction during non-random pointer orientations was 3.4 mm, occurring in the Y direction during a change in pitch. Changing from pitch position 1 to pitch position 2 resulted in the largest deviations in the X, Y and Z directions of all the non-random orientations. This could be because pitch position 1 → pitch position 2 was the largest change in pitch with a difference of 30.7° between positions compared to 13.4° and 9.3° for the following orientations. However, this doesn't explain why the deviations were larger in pitch when compared to roll and yaw. The maximum change in roll orientation was 30°, resulting in a maximum marker deviation of 1.03

mm and the maximum change in yaw was 23°, resulting in a maximum marker deviation of 0.63 mm.

The smallest mean absolute error for a non-random orientation was 0.44 mm, occurring in yaw and the largest was 1.69, occurring in pitch. These results all suggest that marker position is most sensitive to changes in pitch and least sensitive to changes in yaw. This suggests that the larger deviations seen when moving from random orientation position 1 to 2 are more likely to be due to a change in pitch than a change in roll or yaw.

These results suggest that the maximum deviation that could occur in one direction is 5.4 mm. It was determined that a deviation of this magnitude from the true anatomical landmark position could have a small but significant effect on kinematic output. There was no significant difference in flexion/extension or internal/external rotation for the 4 different marker positions, however there was a significant difference in ab/adduction. T-tests for each marker position revealed that there was a significant difference in ab/adduction when the medial epicondyle marker was moved 5.4 mm in the anterior, posterior and distal directions. This is likely to be due to misalignment of the flexion axis caused by relocation of one of the markers which defines it. There is extensive evidence to suggest that misalignment of the flexion axis can result in crosstalk between flexion/extension and ab/adduction, which agrees with results seen in the current study (Della Croce et al., 1999; Fukaya et al., 2013; Morton et al., 2007; Stagni et al., 2006).

Overall, these results suggest that large changes in pitch of the pointer could result in a small deviation of marker position which could give a significantly different knee kinematic output, particularly for ab/adduction in swing phase. However, a number

of suggestions can be made in order to limit the chance of such an error when using a pointer to locate anatomical landmarks. Large changes in pointer pitch should be avoided when placing the tip of the pointer on the anatomical landmark. If pitch is necessary, the pointer should be orientated upwards. If the pointer is orientated downwards, any error which may occur is likely to move the marker in the negative Y direction (distally). This should be avoided as distal deviation may lead to a significant difference in kinematic output. Changes in roll and yaw may be more acceptable as they are less likely to cause large errors in marker position.

There were some limitations to this test. Although a mark was made on a secure surface, it may be possible that the tip of the pointer was not placed in exactly the same position each time a measurement was made. Further, the tip of the pointer was not rigidly attached to the mark during changes in orientation and therefore it is possible that it may have moved slightly when the orientation was altered. The mean camera residual was 0.99 mm which suggests marker positions could be subject to up to 0.9 mm of error. An error in marker position of marker 1 or 2 on the pointer would have an effect on the vector extension algorithm and therefore affect the position of the pointer tip. It is possible that small errors in pointer marker location could have contributed to the deviations recorded in the measured marker position.

In conclusion, the maximum deviation from true marker position that may occur in one direction when changing the orientation of the pointer is 5.4 mm and this may have a small but significant effect on abduction. Large changes in pitch are likely to result in larger deviations from the true marker position. Therefore, when placing the pointer on anatomical landmarks, large changes in pitch should be avoided. However, changes in roll and yaw are likely to be more acceptable.

3.9 Participant Calibration Sequence

The following section details the sequence of events which occurs during calibration of a segment. The code for subject calibration is presented in appendix 1. The user palpates the anatomical landmark of interest and uses the pointer to store its position and the position of the 4 corresponding cluster markers for the segment. When all anatomical landmarks for a segment have been stored, the joint centre, ARF and TRF are calculated. Calculation of the TRF produces a matrix which can be used to describe points within this reference frame. Multiplying the position of an anatomical landmark by the TRF will result in the point being described within the TRF. Therefore, this matrix is stored during calibration, to allow description of anatomical landmarks with respect to the cluster during dynamic trials. The maths required to achieve this is described in the following sections.

To calculated the HJC, the Bell (1989) regression equation was used in this study as evidence from section 3.5.1 suggests it is the one of the more accurate predictive methods for estimation of HJC location. First, pelvic geometry and anatomical landmarks were described in the pelvic ARF by multiplying them by the pelvic G to A rotation matrix (chapter 2). Equation 3.2 through Equation 3.4 were then applied to give an estimation of the location of the right HJC in the pelvic ARF.

$$HJCx = RASISx + (-0.24 * pelvis\ width)$$

Equation 3.2

$$HJCy = RASISy + (-0.30 * pelvis\ width)$$

Equation 3.3

$$HJCz = RASISz + (0.14 * pelvis\ width)$$

Equation 3.4

For the left HJC the calculations are the same except for Equation 3.4 which uses -0.14 instead of 0.14. As described in chapter 2, estimation of the KJC and AJC was calculated by taking the midpoint of the ME and LE and MM and LM, respectively.

The ARF is calculated as described in chapter 2 and from the ARF the fixed calibration anatomical to technical matrix is calculated. This is done by describing the cluster markers in the ARF relative to the technical origin where P^G is a point described in the GRF, ${P^G}_t$ is a point described in the GRF with technical origin, O_t is the technical origin, ${P^A}_t$ is a point described in the ARF and *G to A* is the global to anatomical rotation matrix (Equation 3.5 and Equation 3.6; Figure 3.18).

$$\begin{bmatrix} x \\ y \\ z \end{bmatrix}_{{P^G}_t} = \begin{bmatrix} x \\ y \\ z \end{bmatrix}_{P^G} - \begin{bmatrix} x \\ y \\ z \end{bmatrix}_{O_t}$$

Equation 3.5

$$\begin{bmatrix} x \\ y \\ z \end{bmatrix}_{{P^A}_t} = \begin{bmatrix} Xx & Yx & Zx \\ Xy & Yy & Zy \\ Xz & Yz & Zz \end{bmatrix}_{G\ to\ A} \begin{bmatrix} x \\ y \\ z \end{bmatrix}_{{P^G}_t}$$

Equation 3.6

All cluster markers and anatomical landmarks are described in the ARF with the technical origin in this manner. From these points, the TRF is constructed as described in chapter 2. The 3x3 matrix generated is the fixed anatomical to technical (A to T^F) rotation matrix which can be used to transform a point from the ARF to the TRF. The A to T matrix and anatomical landmarks described in the ARF are used during dynamic trials to reconstruct the position of anatomical points relative to the position of the associated cluster.

3.10 Tracking Segments during Dynamic Trials

When using a cluster based marker set, the A to T matrix and the positions of the anatomical landmarks in the static ARF generated in the calibration stage are

essential for reconstruction of the position of the anatomical landmarks as virtual markers during dynamic trials.

During dynamic trials, a tracking TRF is created within the GRF for each cluster as described in chapter 2, resulting in a technical to global (T to G) rotation matrix. For each segment, a fixed A to T matrix and a dynamic T to G matrix now exists (A to T^F and T to G^D, respectively). These two matrices are multiplied, resulting in the dynamic anatomical to global rotation matrix (A to G^D; Equation 3.7).

$$\begin{bmatrix} Xx & Yx & Zx \\ Xy & Yy & Zy \\ Xz & Yz & Zz \end{bmatrix}_{A\ to\ G^D} = \begin{bmatrix} Xx & Yx & Zx \\ Xy & Yy & Zy \\ Xz & Yz & Zz \end{bmatrix}_{A\ to\ T^F} \begin{bmatrix} Xx & Yx & Zx \\ Xy & Yy & Zy \\ Xz & Yz & Zz \end{bmatrix}_{T\ to\ G^D}$$

Equation 3.7

The *A to* G^D matrix is then used to transform points from the ARF to the GRF where $P^G{}_t$ is a point described in the GRF with global origin and $P^A{}_t$ is a point described in the ARF with technical origin (Equation 3.8).

$$\begin{bmatrix} x \\ y \\ z \end{bmatrix}_{P^{G}{}_t} = \begin{bmatrix} Xx & Yx & Zx \\ Xy & Yy & Zy \\ Xz & Yz & Zz \end{bmatrix}_{A\ to\ G^D} \begin{bmatrix} x \\ y \\ z \end{bmatrix}_{P^{A}{}_t}$$

Equation 3.8

Equation 3.8 gives a point described in the GRF with technical origin. However, in order to reconstruct anatomical landmarks for each segment, the points must be described in the GRF with the global origin. Therefore, the final step in the process is to add the global, dynamic position of the origin of the TRF to $P^G{}_t$ in order to have a point described in the GRF with global origin (Equation 3.9).

$$\begin{bmatrix} x \\ y \\ z \end{bmatrix}_{P^{G}{}_g} = \begin{bmatrix} x \\ y \\ z \end{bmatrix}_{P^{G}{}_t} + \begin{bmatrix} x \\ y \\ z \end{bmatrix}_{O_t}$$

Equation 3.9

For each segment, anatomical landmarks are described in the GRF from the position of the cluster. These points can therefore be reconstructed and used to calculate joint

centres and ARFs for kinematic calculation during dynamic trials. Figure 3.18 is a graphical representation of transforming points from one reference frame to another.

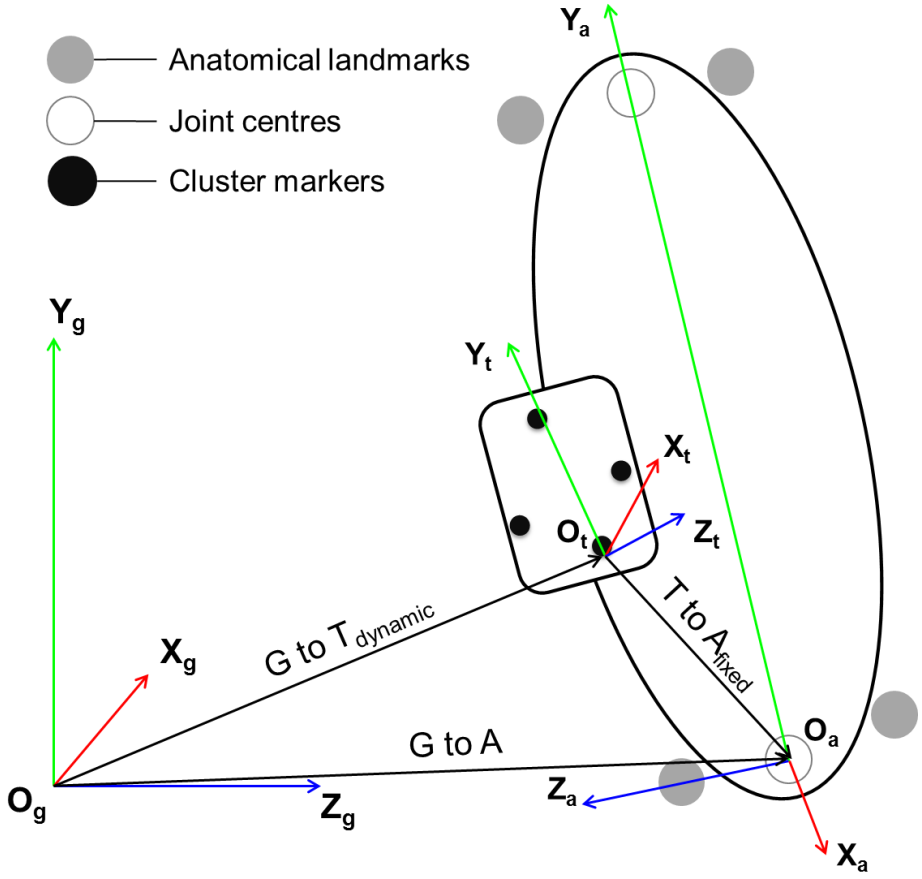


Figure 3.18 Example of transforming points from the GRF and TRF to ARF using the matrices derived in Equation 3.6 through Equation 3.8 where O indicates the origin, subscript g indicates GRF, subscript t indicates TRF and subscript a indicates ARF. G to $T_{dyanmic}$ is the global to technical matrix and can be used to transform points from the GRF to the TRF. T to A_{fixed} is the technical to anatomical matrix and can be used to transform points from the TRF to the ARF. G to A is the product of G to $T_{dyanmic}$ and T to A_{fixed} and can be used to transform points from the GRF to the ARF.

Once anatomical landmarks are reconstructed for each segment, joint centres are calculated using the methods described in section 3.9 and dynamic ARFs are devised for each segment using the methods described in chapter 2. Dynamic ARFs are then used to calculate kinematics.

3.11 Calculation of Kinematics

Sutherland (2002) stated that the calculation of individual joint angles has been recognised as an essential measurement requirement. As a result of this, observation and measurement of kinematics is a primary factor in modern motion analysis. Kinematics may be defined as measurement of the way the body moves (Baker, 2013) and generally, the primary focus of kinematic calculation is quantification of joint angles. A clinical description of joint angle consists of three components, one in each anatomical plane of the body (Figure 3.19; Cole et al., 1993).

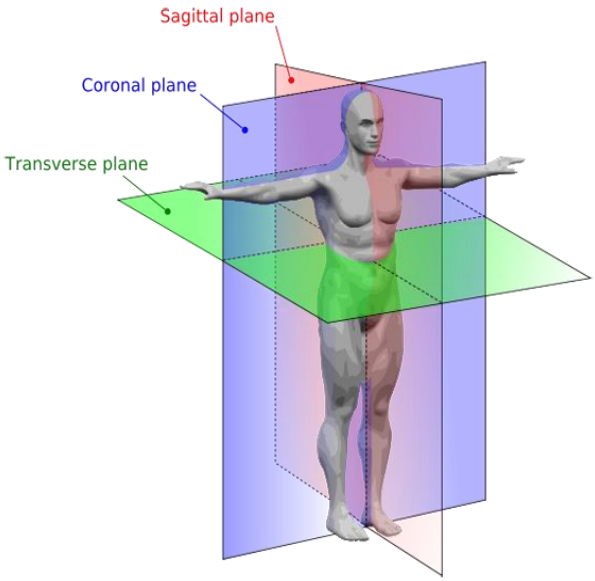


Figure 3.19 Anatomical planes of the body

For each joint, there must be a moving segment and a reference segment, about which the measured segment moves. Generally, the reference segment is the segment proximal to the joint of interest and the moving segment is the segment distal to the joint of interest. Flexion/extension is described as rotation of the distal segment in the sagittal plane of the proximal segment. Abduction/adduction is described as rotation of the distal segment away from or towards the sagittal plane

(i.e. in the coronal plane) and internal/external rotation is defined as rotation of the distal segment about its long axis (i.e. in the transverse plane).

Euler angles can be used to describe the rotation of a distal segment with respect to a proximal segment in 3D space. Euler angles are defined as a set of 3 rotations which take place in sequence to achieve a final orientation from a reference orientation (Kadaba et al., 1990). However, the resultant angles are dependent on rotation order and literature suggests a number of differing conventions for Euler angles are in use (Grood and Suntay, 1983). Dependency on rotation order is a result of 3 separate rotations occurring about the axes of the reference frame located in the moving body. Equation 3.10 through Equation 3.12 highlight the issue with dependency on rotation order, where r_1 is a vector described in reference frame 1, r_2 is the same vector expressed in reference frame 2 and $[R_\alpha]$, $[R_\beta]$ and $[R_\gamma]$ are an ordered set of rotations performed about the axes of the moving reference frame 1. If the order of α and β are reversed, the result will be a different displacement (Equation 3.10 through Equation 3.12; Grood and Suntay, 1983).

$$r_2 = [R_\alpha][R_\beta][R_\gamma]r_1$$

Equation 3.10

$$r_2' = [R_\beta][R_\alpha][R_\gamma]r_1$$

Equation 3.11

$$[R_\alpha][R_\beta] \neq [R_\beta][R_\alpha]$$

Equation 3.12

Grood and Suntay (1983) proposed a new method of kinematic calculation which employs a 'floating' axis for each joint which hides the dependency on rotation order and defines angles in a clinically relevant way. In order to calculate kinematics for a joint, a non-orthogonal 'working' axis system is defined for each joint. This involves taking the cross product of the medio-lateral axis of the proximal segment and the

long axis of the distal segment to create the floating axis. These 3 axes are then employed as the working axis system for the joint (Figure 3.20) where rotation about each axis describes a clinically relevant movement. Rotation about the proximal medio-lateral axis describes the flexion/extension angle (α). Rotation about the distal long axis describes the internal/external rotation angle (β) and rotation about the floating axis describes the abduction/adduction angle (γ).

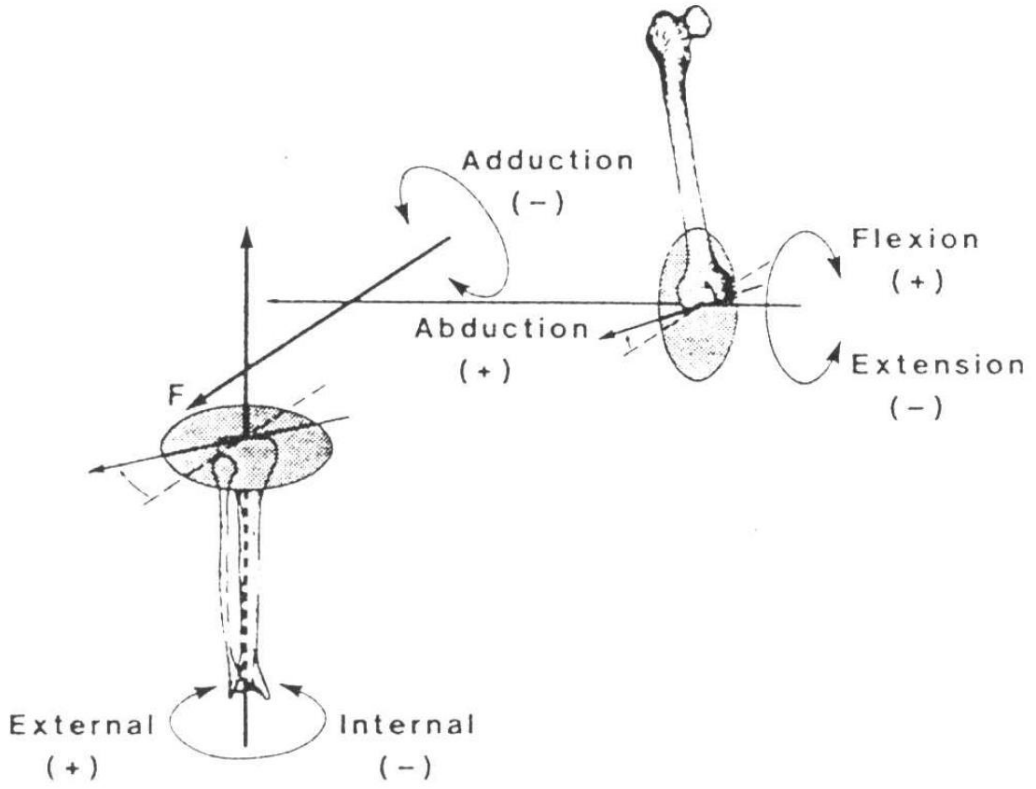


Figure 3.20 Working axis system for the right knee as described by Grood and Suntay, (1983) where F is the floating axis

This method was implemented in this study and involves creating a set of 3 axes termed the 'joint axis system' for each joint, about which the above described movements can occur. The unit vectors of axes are denoted as e_1, e_2 and e_3 where two of the axes are described as body fixed axes and are embedded in the segments whose movement is to be described. e_1 is the proximal segment body fixed axis and

e_3 is the distal segment body fixed axis. The third axis is termed the floating axis because it is not fixed to either segment and moves in relation to both. The floating axis is denoted as e_2 and is mutually perpendicular to e_1 and e_3. Therefore, its orientation is given by the cross product of the unit vectors of the body fixed axes (Equation 3.13).

$$e_2 = \frac{e_1 \times e_3}{|e_1 \times e_3|}$$

Equation 3.13

When first introduced, this method was applied to the knee joint (Grood and Suntay, 1983). However, Cole et al., (1993) proposed a generalised algorithm which could be applicable to all joints in the lower limb, regardless of the way in which ARFs were defined. It was proposed that each segment should have a set of axes as outlined in Table 3.10.

Table 3.10 Axis definitions and nomenclature for the generalised algorithm for calculation of joint kinematics as proposed by Cole et al., (1993)

Name	Abbreviation	Orientation	Unit Vector
Axis of flexion	F-axis	Medio-lateral direction	$\hat{f}$
Longitudinal axis	L-Axis	Predominantly lengthwise	$\hat{l}$
Third axis	T-Axis	Cross product of F-axis and L-axis	$\hat{t}$

Figure 3.21 is an example of the axis definitions proposed by Cole applied to the right thigh. As outlined in chapter 2, all ARFs in this model are created in the same way therefore the F, L and T axes will always correspond to the same X, Y and Z axes for each segment.

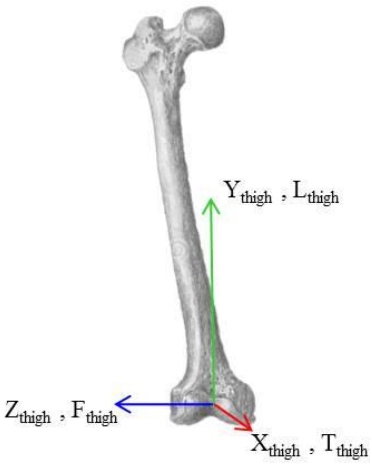
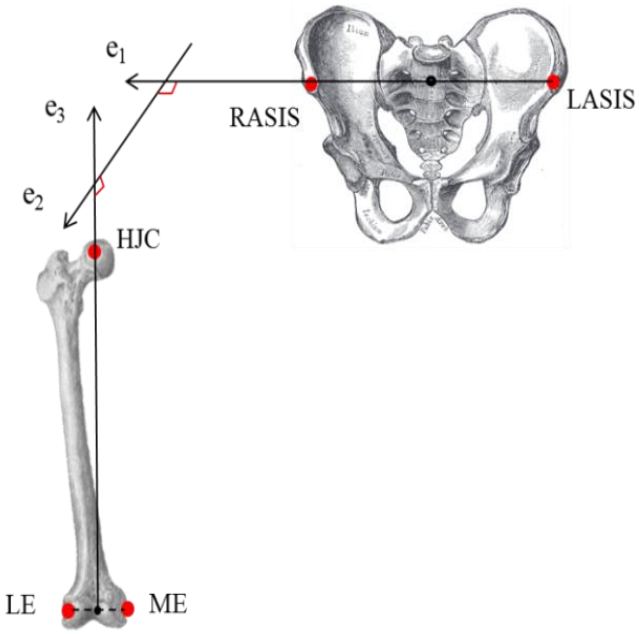


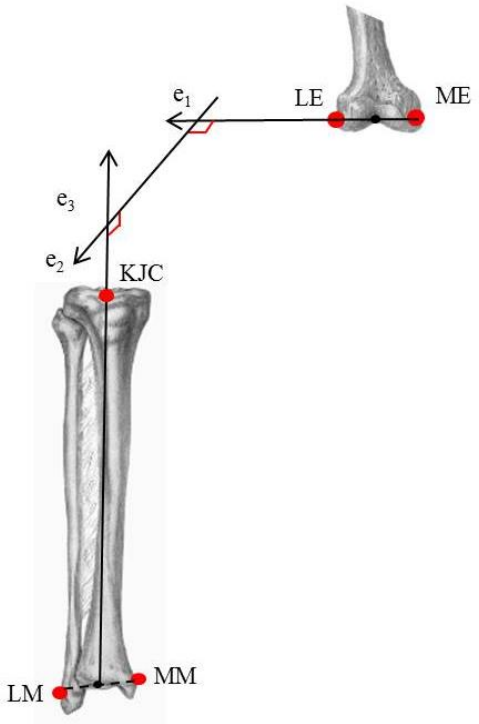
Figure 3.21 X, Y and Z axes of the thigh as described by Cole's generalised algorithm. L – longitudinal axis, F – Flexion axis, T – Third axis

When constructing the joint axis system for any joint, it is necessary to specify the fixed coordinate system in each segment. It is also necessary to describe the body fixed axes of the joint coordinate system and the reference axes which will be used to describe relative motion. It is often convenient to specify the segment ARF in such a way that the existing axes can be used to define the joint coordinate system and reference axes. With this in mind, for each joint, the medio-lateral axis of the proximal segment is used as the proximal body fixed axis (e_3) and the long axis of the distal segment is used as the distal body fixed axis (e_1). Figure 3.22 through Figure 3.24 describe definition of the joint axis systems for the hip, knee and ankle, respectively.



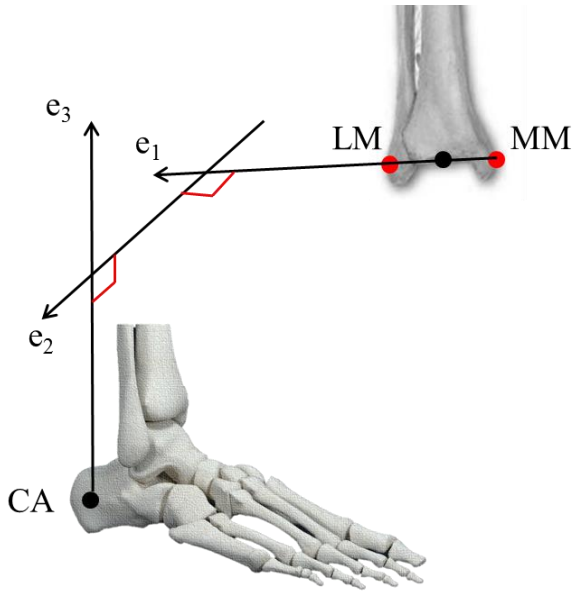
Proximal body fixed axis	e_1 - z axis of the pelvis
Distal body fixed axis	e_3 - y axis of the femur
Clinical description of movement	Rotation of the femur (distal segment) about e_1 describes flexion/extension of the hip. Rotation about e_2 describes abduction/adduction and rotation about e_3 describes internal/external rotation.

Figure 3.22 Joint axis system for the hip. RASIS – right anterior superior iliac spine, LASIS – left anterior superior iliac spine, HJC – hip joint centre, LE – lateral epicondyle, ME – medial epicondyle



Proximal body fixed axis	e_1 – z axis of the femur
Distal body fixed axis	e_3 – y axis of the tibia
Clinical description of movement	Rotation of the tibia (distal segment) about e_1 describes flexion/extension of the knee. Rotation about e_2 describes abduction/adduction and rotation about e_3 describes internal/external rotation.

Figure 3.23 Joint axis system for the knee. LE – lateral epicondyle, ME – medial epicondyle, LM – lateral malleolus, MM – medial malleolus, KJC – knee joint centre



Proximal body fixed axis	e_1 – z axis of the tibia
Distal body fixed axis	e_3 – y axis of the foot
Clinical description of movement	Rotation of the foot (distal segment) about e_1 describes plantar/dorsi flexion of the ankle. Rotation about e_2 describes abduction/adduction and rotation about e_3 describes internal/external rotation.

Figure 3.24 Joint axis system for the ankle. LM – lateral malleolus, MM – medial malleolus, AJC – ankle joint centre

As previously mentioned, rotation about each axis describes a clinically relevant movement. However, the magnitude of rotation about each body fixed axis cannot be measured relative to any of the other axes in the working axis system. Therefore, reference axes (e^r) are required in order to quantify rotation about each body fixed axis. The reference axis is defined as the axis perpendicular to the axis about which the rotation is occurring.

For example, considering the right knee, rotation about the femoral body fixed axis would produce a flexion/extension movement. This cannot be quantified in relation to any of the other axes in the working axis system. Therefore, the femoral X axis is employed as the flexion/extension reference axis (e_1^r) and the angle between (e_1^r) and (e_2) represents the flexion/extension angle (α; Figure 3.25a).

Continuing with the example of the right knee, rotation about the tibia body fixed axis would produce an internal/external rotation movement. This can also not be quantified in relation to the other axes in the working axis system. Therefore, the tibia X axis is employed as the internal/external rotation reference axis (e_3^r) and the angle between e_3^r and the floating (e_2) axis represents the internal/external rotation angle (γ; Figure 3.25b).

a. **b.**

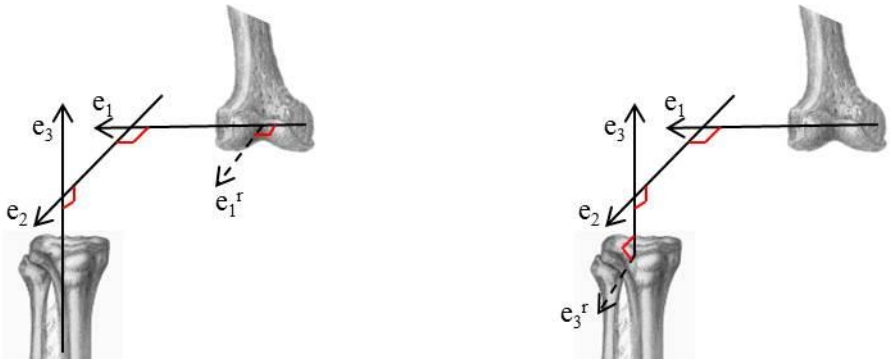


Figure 3.25 Reference axes for movements not quantified by use of the working axis system **a.** Flexion/extension reference axis **b.** Internal/external rotation reference axis

The abduction/adduction angle (β) is calculated as the angle between the two body fixed axes. Equation 3.14 through Equation 3.16 describe calculation of each clinical movement, where subscript i refers to axes which are part of the proximal segment, subscript j refers to axes which are part of the distal segment and subscript ij refers to axes which are part of the working axis system.

Flexion/extension is calculated using Equation 3.14 where e_{2ij} is the floating axis, t_i is the third axis of the proximal segment i.e. the flexion/extension reference axis (e_1^r) and l_i is the proximal longitudinal axis. B determines the sign of the angle, which is positive if the angle between the floating axis and the proximal longitudinal axis is greater than zero and negative otherwise.

$$\sigma = cos^{-1}\left(e_{2_{ij}} \cdot t_i\right) \times B$$
where $B = 1\ if\ \left(e_{2_{ij}} \cdot l_i\right) > 0\ else\ B = -1$

Equation 3.14

Abduction/adduction is calculated using Equation 3.15 where r is an axis mutually orthogonal to the proximal flexion axis (f_i) and the floating axis and orientated downwards and l_j is the distal longitudinal axis. C determines the sign of the angle and is positive if the angle between the proximal flexion axis and the distal longitudinal axis is greater than zero and negative otherwise.

$$\beta = cos^{-1}\left(r \cdot l_j\right) \times C$$
where $r = \left(\frac{f_{i \times e_{2_{ij}}}}{\left|f_i \times e_{2_{ij}}\right|}\right)$
and $C = 1\ if\ \left(f_i \cdot l_j\right) > 0\ else\ C = -1$

Equation 3.15

Internal/external rotation is calculated using Equation 3.16 where t_j is the third axis of the distal segment and f_j is the flexion axis of the distal segment. D determines the sign of the angle and is positive if the angle between the floating axis and the distal flexion axis is greater than zero and negative otherwise.

$$\gamma = cos^{-1}\left(e_{2_{ij}} \cdot t_j\right) \times D$$
where $D = 1\ if\ \left(e_{2_{ij}} \cdot f_j\right) > 0\ else\ D = -1$

Equation 3.16

The code which was used to track segments and calculate kinematics is presented in appendix 1.

Chapter 4: Comparison of Strathclyde Cluster Model to Plug in Gait

Chapter 4

4 Comparison of Strathclyde Cluster Model to Plug in Gait

4.1 Introduction

As evidenced in section 3.6, differences in biomechanical models can result in differences in kinematic output. As the model proposed here is bespoke and was custom written, a comparison was carried out between SCM and the current clinical gold standard, PiG, to determine if the kinematic output calculated by SCM was acceptable for clinical use. Limited work has been carried out to investigate differences between cluster based models and models which employ skin surface markers. However, evidence presented in chapter 3 suggests that cluster based models are subject to reduced STA when compared to skin surface marker models. As a result of this, there are likely to be some differences between models. However, without comparison to a bone fixated device, it is not possible to determine which model is producing the most accurate measure of kinematics. Therefore the purpose of this comparison is not to validate SCM, rather it is to assure the clinical community that a bespoke, cluster based model is capable of producing a meaningful kinematic output.

4.1.1 Methods

This investigation was approved by the departmental ethics committee at the Department of Biomedical Engineering, University of Strathclyde. Five participants volunteered for the study which took place in the Biomechanics Lab in the Department of Biomedical Engineering, University of Strathclyde. Inclusion criteria for the study required all participants to be able-bodied, have normal lower limb function and be able to walk at a self-determined pace for approximately 500m

without excess physical exertion or pain. The same 12 camera Vicon system described in section 3.8.1 was used in this study. Prior to testing each participant the system was calibrated using the methods described in chapter 2. A comprehensive marker set was designed, allowing participants to wear both SCM and PiG simultaneously (Figure 4.1).

Figure 4.1 Comprehensive marker set comprised of SCM and PiG

Participants were asked to wear tight clothing to allow placement of skin surface markers to be as accurate as possible and markers were applied by the same researcher for each participant to minimise inter-rater variability. This study took place before a pointer based calibration method had been developed. Therefore, participant calibration for SCM was carried out using the static method which employs a number of skin surface markers. The location of all markers of the comprehensive marker set is detailed in Figure 4.1 and Table 4.1.

Table 4.1 Marker locations for comprehensive marker set

Segment	Marker	Marker Location	Model
Pelvis	RASIS	Over the right anterior superior iliac spine	PiG/SCM (calibration only)
	LASIS	Over the left anterior superior iliac spine	PiG/SCM (calibration only)
	RPSIS	Over the right posterior superior iliac spine	PiG/SCM (calibration only)
	LPSIS	Over the left posterior superior iliac spine	PiG/SCM (calibration only)
	Pelvis cluster	On the back, below the PSIS markers	SCM
Thigh	LTHI	Approximately 1/3 of the distance between then greater trochanter and the LME on the lateral surface of the thigh	PiG
	RTHI	Approximately 2/3 of the distance between then greater trochanter and the RME on the lateral surface of the thigh	PiG
	RME/LME	Over the medial epicondyle of the right and left knee	SCM (calibration only)
	RLE/LLE	Over the lateral epicondyle of the right and left knee	PiG/SCM (calibration only)
	L thigh cluster	On the lateral surface of the thigh, below LTHI	SCM
	R thigh cluster	On the lateral surface of the thigh, above RTHI	SCM
Shank	LTIB	Approximately 1/3 of the distance between then LME and the LMM on the lateral surface of the shank	PiG
	RTIB	Approximately 2/3 of the distance between then RME and the RMM on the lateral surface of the shank	PiG
	RMM/LMM	Over the medial malleolus of the right and left ankle	SCM (calibration only)
	RLM/LLM	Over the lateral malleolus of the right and left ankle	PiG/SCM (calibration only)
	R shank cluster	On the antero-lateral surface of the shank, below RTIB	SCM
	L shank cluster	On the antero-lateral surface of the shank, below LTIB	SCM
Foot	RHEE/LHEE	Over the calcaneus of the right and left foot	PiG/SCM
	RFM/LFM	Over the dorsal surface of the distal end of the right and left first metatarsal	PiG/SCM
	RVM/LVM	Over the dorsal surface of the distal end of the right and left fifth metatarsal	SCM

Skin surface markers were attached with double-sided tape and the clusters were attached using Velcro straps, in the positions described in section 3.7. Clusters were secured to minimise movement and avoid slippage, but also to maximise comfort for the wearer.

A static calibration trial was captured which allowed calibration of PiG and SCM. Anthropometric measurements required by PiG were also calculated at this point. These included interASIS distance, knee width, ankle width and leg length. These measurements were taken using markers located over anatomical landmarks. Table 4.2 describes which markers were used to calculate each anthropometric measurement. This was done for right and left legs. Participant height was also measured.

Table 4.2 Anthropometric measurements and markers used to calculate them

Anthropometric Measurement	**Markers Used**
interASIS distance	RASIS and LASIS
Knee width	LE and ME
Ankle width	LM and MM
Leg length	ASIS and MM

Following calibration, medial SCM calibration markers were removed to prevent them from compromising participant movement. Participants were then asked to perform a number of practice walks to allow them to become familiar with moving whilst wearing the markers. Following familiarisation, a minimum of 10 walking trials were captured. All participants were shod and performed walking trials at a self-selected speed.

Trial data for each model was processed using the respective methods required for each marker set. From each walking trial, one trial containing the PiG marker set and one trial containing the SCM marker set was created. PiG data were processed

using the standard dynamic PiG pipeline in the Vicon Software (Nexus). Marker trajectories were filtered using a 4th order Butterworth filter with a cut off frequency of 10hz and smoothed. The dynamic PiG model was then run and an ASCII output was generated. SCM data were processed using D-Flow. Prior to processing, markers were labelled and any gaps were filled using Nexus. Trajectories were then streamed into D-Flow and filtered using a second order Butterworth filter with a cut off frequency of 10 Hz. Trials were processed using bespoke code (appendix 1) and the methods described in sections 3.9 through 3.11.

Data analysis was performed using MATLAB (Mathworks Inc). Data from all PiG and SCM trials were stacked and normalised to 100% of the gait cycle using a linear regression fit (appendix 2). Comparisons were made between flexion/extension, abduction/adduction and internal/external rotation for the right and left hip and knee. Right and left ankle dorsi/plantar flexion was also compared. A running paired t – test ($\alpha = 0.05$) was carried out at each percent of the gait cycle to determine any areas of significant difference between models. A number of parameters were also compared at distinct stages of the gait cycle for a representative participant and for all participants. Further, the mean total joint excursion for all participants was compared.

4.1.2 Results

All gait cycles from a representative subject were processed using PiG and SCM with data from the right leg presented below. Figure 4.2 shows mean kinematics ± 2 standard deviations (SD) for PiG and SCM outputs for the right leg of one subject over 8 walking trials. On the graphs in this section and all subsequent kinematic graphs, knee and hip flexion and internal rotation are positive, hip adduction is

positive and ankle dorsiflexion is positive. For the knee, abduction (varus) is positive, which actually corresponds to shank adduction, as it causes the distal segment to be adducted with respect to the proximal segment. Therefore knee abduction is equal to shank adduction.

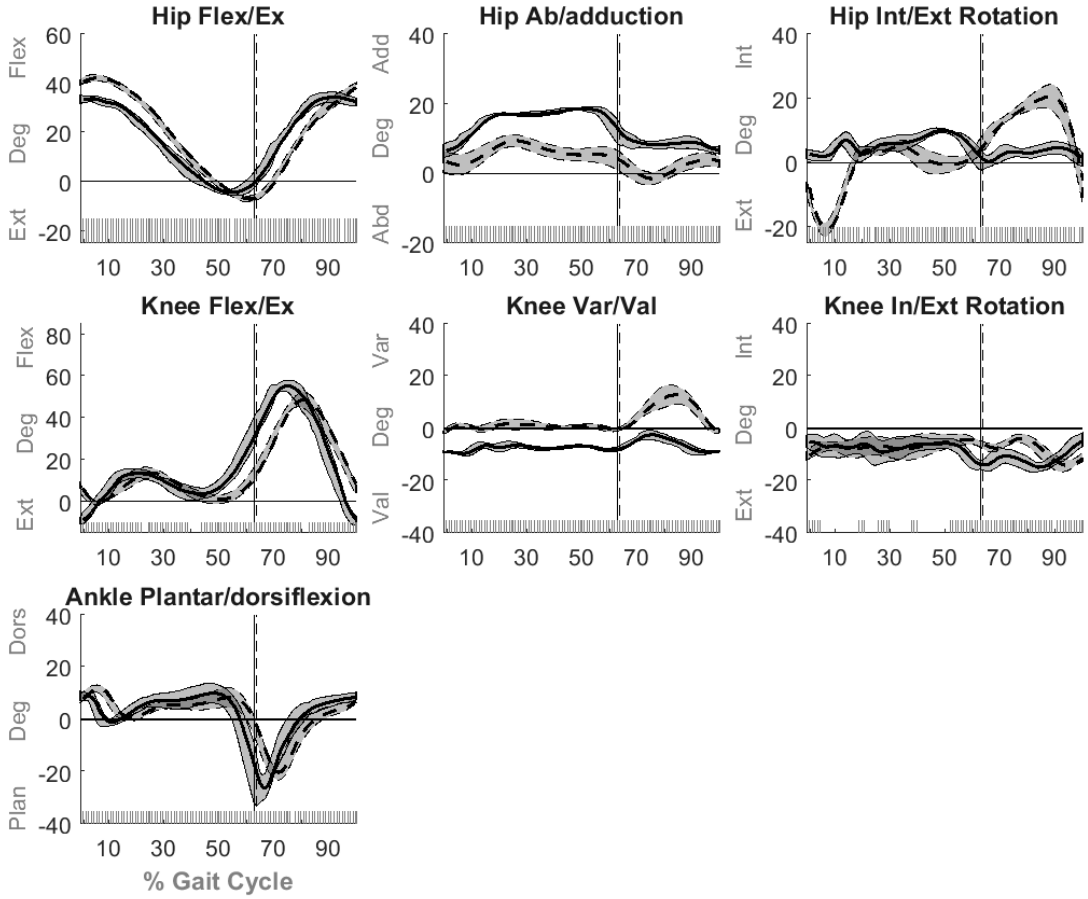


Figure 4.2 Mean kinematic output for one subject's right leg over 8 walking trials. Shaded grey areas represent mean ± 2SDs. Grey bars represent areas of significant difference (α = 0.05). Mean toe off is represented by vertical lines. PiG (dashed) SCM (solid)

There were some areas of significant difference for all rotations at all joints as indicated by the stippling along the horizontal axis (Figure 4.2). Best agreement was observed in knee internal/external rotation. Least agreement was observed in hip adduction and knee abduction as there was an offset of approximately 10°, although the excursion appeared similar. For hip flexion there was a significant difference for almost the entire gait cycle, however the excursion appeared similar for the two

models. For hip adduction, SCM produced larger angles in comparison to PiG however PiG was more variable. There was some agreement in excursion and trend between models for hip adduction. For hip internal/external rotation there was little agreement in ROM, excursion or shape of the curve. For knee flexion, the shape of the curve and excursion were similar in early and mid-stance, however there were still some significant differences. During late stance and early swing, SCM showed more flexion in comparison to PiG, whereas in late swing SCM showed less flexion in comparison to PiG. As a result of this, there was a significant difference between models for the majority of swing. Variability in knee flexion was similar for both models. For knee abduction, SCM consistently showed less abduction in comparison PiG and therefore there was a significant difference between models for the whole gait cycle. Excursion in stance was similar for both models; however PiG estimated greater excursion in swing and also exhibited greater variability in swing. There was good agreement between models in stance for knee internal/external rotation with trace, excursion and variability exhibiting similarities. However, in late stance and early swing, SCM showed less internal rotation in comparison to PiG and in late swing SCM showed more internal rotation. This resulted in a significant difference between models for the duration of late stance and swing. There was good agreement for the majority of stance for ankle flexion with similarities in trace, excursion and variability. However in late stance and early swing, SCM showed more plantar flexion in comparison to PiG and was more variable when estimating peak plantar flexion which resulted in significant differences in late stance and swing.

Kinematic parameters were investigated at 6 stages of the gait cycle; flexion at HS (FHS), peak flexion in loading response (PFLR), flexion at midstance (FMS), peak flexion during swing (PFS), peak ab/adduction during swing (PAS) and peak rotation in swing (PRS) and a paired-sample t-test ($\alpha = 0.05$) was used to determine any significant difference between models (Table 4.3 through Table 4.5).

Table 4.3 Mean (±SD) of hip parameters as calculated by each model for a representative subject and corresponding t-test results for the hip. *significant difference ($\alpha = 0.05$)

HIP

Parameter (°)	**PiG mean(SD) °**	**SCM mean(SD) °**	**P-Value**
FHS	41.2 (0.9)	33.5 (0.9)	**< 0.001***
PFLR	42.0 (0.9)	33.9 (0.7)	**< 0.001***
FMS	16.6 (0.5)	12.0 (0.6)	**< 0.001***
PFS	39.6 (0.8)	34.5 (1.0)	**< 0.001***
PAS (adduction)	4.0 (1.0)	12.6 (1.6)	**< 0.001***
PRS (external)	14.2 (2.3)	0.0 (0.7)	**< 0.001***

Table 4.4 Mean (±SD) of knee parameters as calculated by each model for a representative subject and corresponding t-test results for the knee. *significant difference ($\alpha = 0.05$)

KNEE

Parameter (°)	**PiG mean (SD) °**	**SCM mean (SD) °**	**P-Value**
FHS	2.4 (1.2)	-8.9 (0.8)	**< 0.001***
PFLR	14.0 (1.5)	13.6 (1.2)	0.085
FMS	9.8 (0.3)	9.6 (0.2)	0.317
PFS	48.6 (1.9)	55.5 (1.4)	**< 0.001***
PAS (abduction)	13.4 (2.0)	-2.4 (0.9)	**< 0.001***
PRS (external)	14.3 (1.1)	15.4 (0.7)	**0.047***

Table 4.5 Mean (±SD) of ankle parameters as calculated by each model for a representative subject and corresponding t-test results for the ankle. *significant difference ($\alpha = 0.05$)

ANKLE

Parameter (°)	**PiG mean(SD) °**	**SCM mean (SD) °**	**P-Value**
FHS (dorsi)	9.5 (0.9)	8.9 (0.9)	**0.0163***
PFLR (dorsi)	11.7 (0.9)	9.5 (0.7)	**<0.001***
FMS (dorsi)	6.5 (0.02)	7.6 (0.009)	**<0.001***
PFS (plantar)	20.3 (1.5)	27.0 (2.3)	**<0.001***

For the hip, SCM estimated a lower angle for all parameters except PAS and variability was higher for PiG than SCM for PFLR and PRS. For the knee, there was

no significant difference for PFLR (P = 0.085) or FMS (P = 0.317) however all other parameters exhibited significant differences. Further, variability was generally lower for SCM. For the ankle, there was a significant difference for all parameters with the largest difference occurring at PFS. Variability was also highest for both models at PFS. Figure 4.3 shows mean results for all participants for SCM and PiG for the right leg.

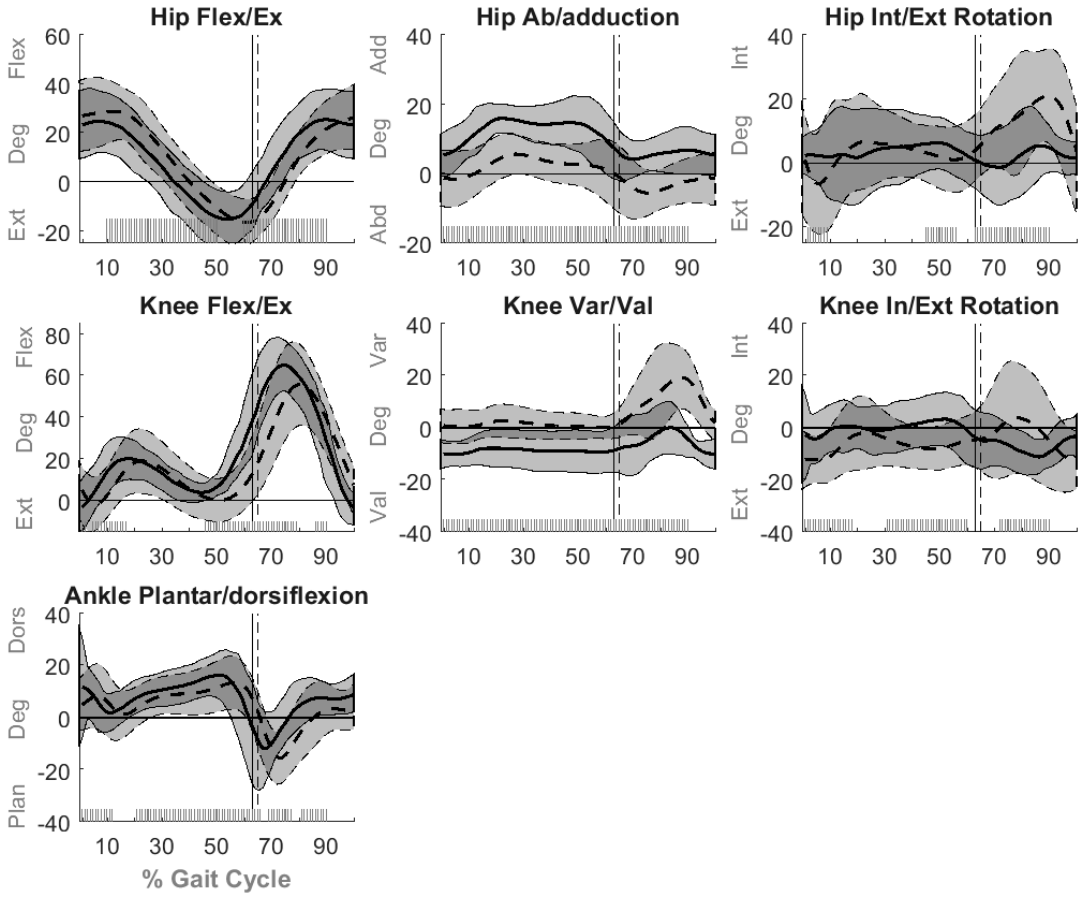


Figure 4.3 Mean kinematic output for all subjects' right legs. Shaded grey areas represent mean ± 2SDs. Grey bars represent areas of significant difference ($\alpha = 0.05$). Mean toe off is represented by solid vertical lines. PiG (dashed) SCM (solid)

When the kinematic outputs from all participants were combined, there were fewer areas of significant difference between models. Further, the variance increased, as would be expected when inter-subject differences are included. However, joint rotations continued to exhibit significant differences, similar to those reported

previously for a representative subject, and again, there were significant offsets present for hip and knee ab/adduction. For hip flexion there were significant differences around mid-stance and early-mid swing. The excursion was similar for both models although SCM showed reduced flexion in stance and slightly higher flexion in swing, compared to PiG. For hip adduction, there was an offset of approximately 10°, as was seen for the representative participant, with SCM consistently estimating more adduction in comparison to PiG. This resulted in a significant difference for the whole gait cycle except late swing. However, the excursion and variability were similar for both models. For hip internal/external rotation, there were significant differences measured in stance and swing and there was little similarity in the shape of the curves or the excursion between models. The largest difference occurred in early swing with PiG estimating greater internal rotation than SCM and also exhibiting higher variability. For knee flexion, there were significant differences in early stance, late stance and early swing. Generally SCM showed greater flexion in comparison to PiG which is particularly evident in swing where PiG estimated lower flexion values than SCM and displayed higher variability. Despite this, the excursion and shape of the curves are similar for both models. For knee abduction there was an offset of about 10° with SCM consistently showing less abduction in comparison to PiG. This resulted in a significant difference for the whole gait cycle except late swing. Further, PiG estimated greater excursion in swing than SCM and exhibited higher variability. For knee internal/external rotation, there were significant differences in early stance, mid-late stance and mid swing. There was little similarity in the shape of the curve or excursion between models as SCM showed more internal rotation in stance and less

internal rotation in swing in comparison to PiG. Further, PiG displayed higher variability in swing and estimated the knee to be internally rotated, in comparison to SCM which estimated the knee to be externally rotated. For ankle flexion there were significant differences across the entire gait cycle apart from early-mid stance and late swing. SCM showed more dorsiflexion in comparison to PiG for the majority of the gait cycle, most evidently in mid-late stance and swing. However, shape of the curve, excursion and variability for both models were similar, apart from at HS where SCM exhibited higher variability than PiG. Table 4.6 through Table 4.8 detail cycle parameter results and mean joint excursions for all participants.

Table 4.6 Mean (±SD) of hip parameters as calculated by each model for all subjects and corresponding t-test results for the hip. *significant difference ($\alpha = 0.05$)

HIP

Parameter (°)	**PiG mean(SD) °**	**SCM mean(SD) °**	**P-Value**
FHS	27.3 (7.4)	23.3 (7.0)	< **0.001***
PFLR	29.2 (6.7)	25.1 (6.5)	< **0.001***
FMS	16.3 (0.9)	11.8 (0.9)	< **0.001***
PFS	26.9 (6.9)	25.8 (6.2)	0.13
PAS (adduction)	-0.2 (3.1)	9.0 (2.8)	< **0.001***
PRS (external)	-3.4 (7.8)	-2.6 (5.5)	0.689
Flex/Ex Excursion	46 (4.7)	41.9 (4.6)	<**0.001***
Abd/Add Excursion	11.8 (2.2)	14.3 (2.2)	<**0.001***
Int/Ext Rotation Excursion	28.5 (7.4)	13.4 (4.6)	<**0.001***

Table 4.7 Mean (±SD) of knee parameters as calculated by each model for all subjects and corresponding t-test results for the knee. *significant difference ($\alpha = 0.05$)

KNEE

Parameter (°)	**PiG mean (SD) °**	**SCM mean (SD) °**	**P-Value**
FHS	0.9 (6.8)	-3.2 (5.7)	< **0.001***
PFLR	19.4 (7.6)	20.6 (5.1)	0.185
FMS	9.6 (0.4)	9.6 (0.4)	0.928
PFS	56.1 (10.2)	65.7 (6.8)	< **0.001***
PAS (abduction)	20.0 (6.1)	1.0 (4.5)	< **0.001***
PRS (external)	13.7 (5.7)	12.2 (2.8)	0.091
Flex/Ex Excursion	58.2 (5.6)	70.4 (3.5)	<**0.001***
Abd/Add Excursion	21.1 (6.8)	12.4 (5.6)	<**0.001***
Int/Ext Rotation Excursion	18.7 (7.1)	18.4 (7.1)	0.97

Table 4.8 Mean (±SD) of ankle parameters as calculated by each model for all subjects and corresponding t-test results for the ankle. *significant difference ($\alpha = 0.05$)

ANKLE

Parameter (°)	**PiG mean(SD) °**	**SCM mean (SD) °**	**P-Value**
FHS (dorsi)	6.3 (5.5)	11.8 (11.4)	**0.003***
PFLR (dorsi)	10.4 (4.0)	13.8 (10.8)	**0.0412***
FMS (dorsi)	6.6 (0.03)	7.7 (0.03)	**< 0.001***
PFS (plantar)	15.9 (5.2)	12.9 (7.8)	**0.003***
Plantar/dorsiflexion Excursion	30.3 (3.1)	31.3 (10.4)	0.62

For the hip, SCM estimated lower values for all parameters except PAS and variability was larger for PiG for all parameters except FMS, which was the same for both models. There was a significant difference for FHS, PFLR, FMS and PAS and for excursions (P<0.001). For the knee there was a significant difference for FHS, PFS and PAS and for flexion/extension and ab/adduction excursions (P<0.001). Variability was lower for SCM for all parameters except FMS, where it was again the same. For the ankle, SCM estimated higher values for all parameters in comparison to PiG except PFS. Further, SCM displayed higher variability in FHS and PFLR, however variability for FMS and PFS were similar to PiG. Therefore, there were significant differences for all parameters, but no significant difference in overall excursion (P=0.62).

4.1.3 Discussion

The aim of this part of the study was to compare the kinematic output of SCM with PiG. Some significant differences were observed between models and similar differences were seen for representative and group data. However, group variability was higher, as would be expected, and exhibited less areas of significant difference. For parameter values, there were less significant differences and higher variability in group data than individual data.

Some difference in model outputs may be accounted for by the use of different methods for defining ARFs and calculating kinematics. PiG uses the lateral femoral epicondyle and the width of the knee to estimate KJC, suggesting that KJC position is estimated as the distance of half the width of the knee in the medial direction from the position of the lateral marker. However SCM uses medial and lateral femoral epicondyle markers which may account for proximal/distal differences in the medial and lateral condyles of the knee, which PiG does not. This may result in proximal/distal differences in estimation of KJC position. Since the HJC and KJC are used to define the thigh and shank ARFs, this is likely to result in different ARF definitions for PiG and SCM which is then likely to lead to a different kinematic output.

For the knee, PiG estimated lower flexion and higher abduction in swing than SCM. PiG measured up to 30° of abduction which is abnormal for a healthy participant (Ferrari et al., 2008). It may therefore be suggested that kinematic crosstalk is occurring between flexion and ab/adduction in the PiG output. Ferrari et al., (2008) reported up to 35° of knee abduction when using PiG with a participant who had their knee fully restrained from ab/adduction, therefore the value was expected to be zero. It was suspected that the high abduction angles resulted from incorrect alignment of the knee flexion axis resulting in crosstalk between flexion and abduction. This phenomenon has been reported elsewhere regarding PiG data (McGinley et al., 2009) and therefore is likely to be the reason for the higher knee abduction angles measured by PiG in this study. These results are also supported by the knee excursion values with PiG estimating a lower flexion/extension excursion and higher ab/adduction excursion when compared to SCM.

The consistent offset in hip and knee ab/adduction is likely to be because, during calibration, PiG adjusts the position of the ASIS markers based on the participant's interASIS distance. SCM doesn't adjust the ASIS landmarks in this way and therefore this may result in differences in ASIS position between models which will have an effect on HJC location and therefore hip and knee ab/adduction.

The excursion results for hip ab/adduction suggest that the difference in kinematic output is caused by more than just an offset, as there was a significant difference between models. However, the average difference was only 2.5° and is therefore likely to be the result of different kinematic calculation methods.

Internal/external rotation has often been reported as the most variable kinematic output (Ferrari et al., 2008; Holden et al., 1997; Karlsson and Lundberg 1994) which is reflected in the results obtained. Unlike flexion and ab/adduction, there is little similarity in the ROM for internal/external rotation at the hip or knee between models. The biggest differences were observed in swing with PiG estimating greater hip internal rotation than SCM and PiG estimating knee internal rotation when SCM estimated external rotation. In a previous study which compared 5 gait protocols (Ferrari et al., 2008), results showed hip internal rotation of more than 5° for only one of the 5 protocols. One protocol estimated external rotation in swing; however the remaining 3 protocols measured an increase from external to internal rotation of approximately 10°. It may therefore be suggested that hip internal rotation of greater than 10° is abnormally high for healthy individuals and SCM is likely to be obtaining a truer representation of hip internal/external rotation in swing than PiG.

Results from Ferrari et al., (2008) also demonstrate 4 out of 5 protocols estimating knee external rotation in the first half of swing, with only one protocol measuring

internal rotation. Further, Czamara et al., (2015) measured 4° of knee external rotation at the point of maximal knee flexion in swing. This supports the results from Ferrari et al., (2008) who measured between 10° and 20° of external rotation for 4 out of 5 protocols and the results from SCM in this study which estimated an average of 12° external rotation in swing.

Therefore, evidence suggests that normal knee rotational movement patterns in swing result in external rotation rather than internal rotation, suggesting that SCM is again likely to be obtaining a truer representation of knee internal/external rotation in swing.

Evidence has suggested internal/external rotation may be more affected by STA than flexion or ab/adduction (Manal et al., 2000). Further, it has been reported that errors in internal/external rotation angle can be up to as much as the expected ROM, and therefore STA could affect internal/external rotation output so much that it is no longer a true representation of joint rotation (Leardini et al., 2005). Since PiG uses skin surface markers and SCM does not, this may account for the higher variability in internal/external rotation output from PiG.

Collins et al., (2009) compared the kinematic output of a standard HH style marker set to a bespoke, cluster based 6 degrees of freedom set (6DoF). A comprehensive marker set was used (Figure 4.4) and kinematics were measured during treadmill walking.

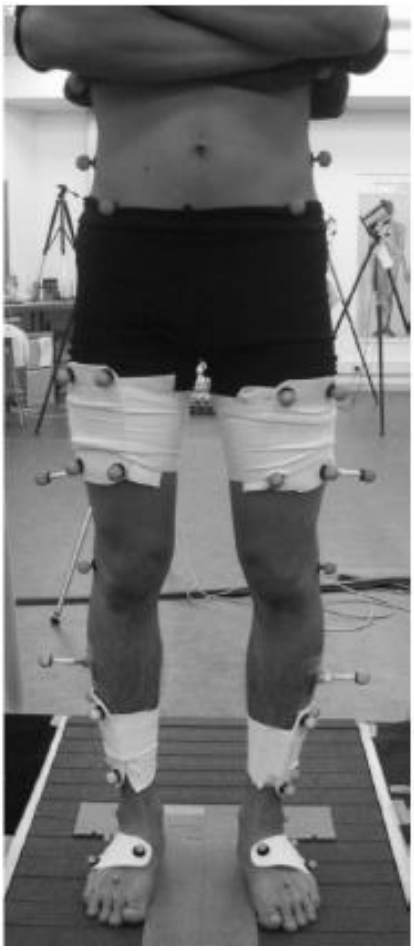


Figure 4.4 Comprehensive marker set allowing comparison of HH protocol and bespoke 6DoF protocol (Collins et al., 2009)

For flexion angles of the hip and knee, both models estimated similar outputs although 6DoF estimated less knee flexion in LR than HH (Figure 4.5). For ankle flexion HH consistently estimated more flexion in comparison to 6DoF and 6DoF was more variable in swing than HH. For hip adduction, HH estimated a similar ROM to 6DoF; however the trace differed from 6DoF, with transition from adduction to abduction occurring earlier in the cycle compared to 6DoF. For knee abduction 6DoF showed greater peak abduction in swing compared to HH, estimating approximately 15° compared to HH which estimated approximately 5°. This result is not supported by the majority of findings regarding HH style marker sets as a number of studies have indicated that abnormally high values are often

reported for knee abduction in swing (Ferrari et al., 2008; McGinley et al., 2009). These results may be due to use of thigh and tibial wands, as opposed to markers, which reduce the chance of error propagation and therefore may result in better alignment of the knee flexion axis (Cappozzo et al., 1995). For knee and hip internal/external rotation there was little similarity between HH and 6DoF. For hip internal/external rotation 6DoF appeared to measure internal rotation where HH measured external and the ROM for HH was greater than for 6DoF. For the knee both models measured external rotation in swing, with 6DoF measuring approximately 15° of external rotation and HH measuring approximately 5°.

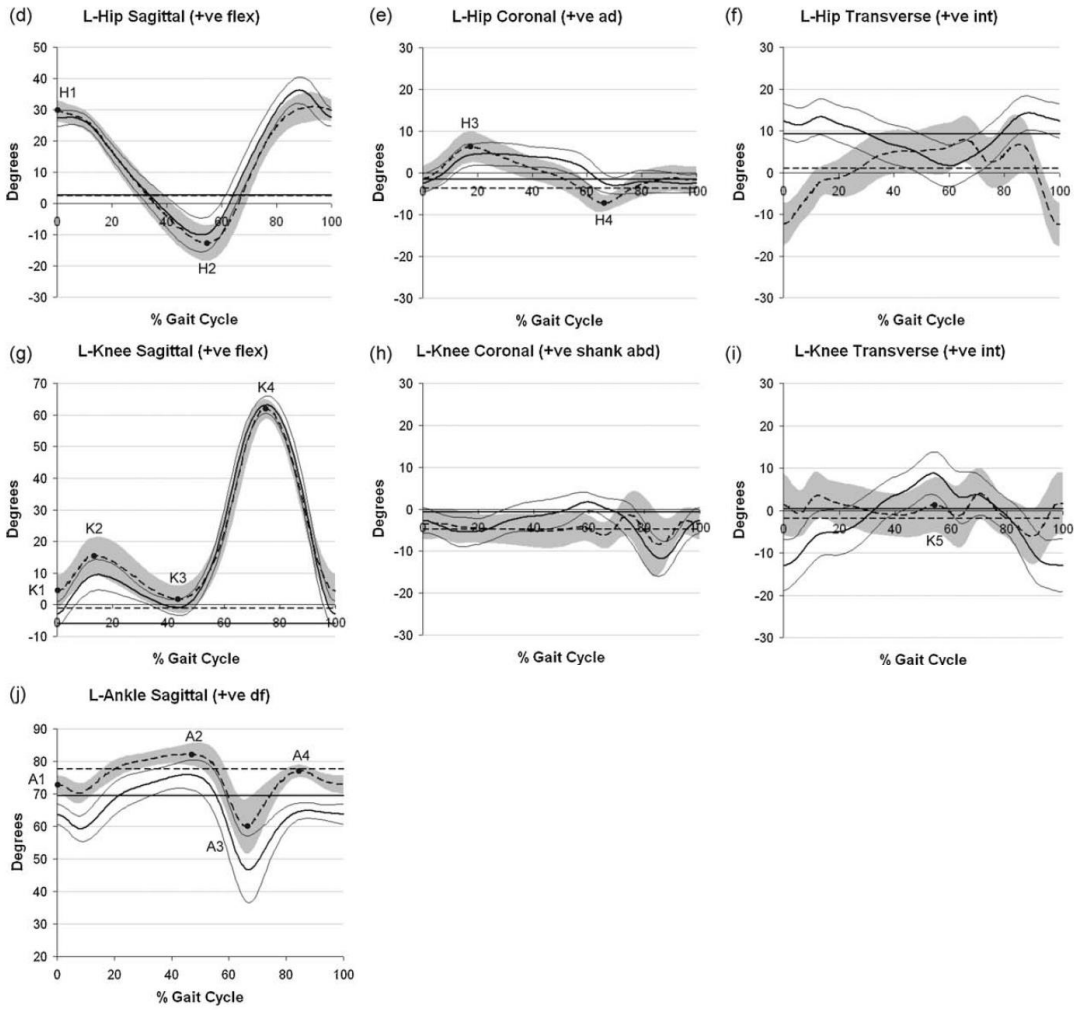


Figure 4.5 Mean kinematic output for 10 subjects for HH (dashed line ± 1SD) and 6DoF (black solid line ± 1SD) (Collins et al., 2009)

However, neither ROM compares to results seen in this study, nor do they compare to results for any of the 5 protocols reported by Ferrari et al., (2008). This further confirms the high variability of internal/external rotation measurements for the hip and knee for different models. The results from the current study are not directly comparable to Collins et al., (2009) as the 6DoF model uses a functional HJC estimation method and there are therefore likely to be differences in HJC location between it and SCM which will have a resultant effect on kinematics. Unlike the results from this study, no offset was observed between hip and knee ab/adduction, it is therefore likely that the 6DoF model also aligns all axes to zero during calibration. HH did not measure more knee abduction in swing and there was less agreement between models for ankle flexion. Further, outputs for internal/external rotation were not only different between HH and 6DoF, but they did not agree with either model used in the current study.

In the current study, for the ankle, there was a significant difference between models for all parameters; however no significant difference between total joint excursions. This suggests that there is a consistent offset throughout the gait cycle in ankle plantar/dorsi flexion angle between models, most likely caused by the different definition of segment ARFs.

One limitation of the current study is that only healthy individuals were tested. PiG is commonly used by a number of clinical laboratories and is relied upon to provide guidance for intervention prescription including surgical planning (Schwartz and Rozumalski, 2005). SCM has so far not been tested on pathological individuals and therefore it cannot presently be recommended for clinical use in this manner. Further, only walking trials were tested and kinematic values for walking are within

certain limits which do not approach the maximum ROM of each joint. The results cannot predict what will happen if more extreme ROMs are tested. PiG is a widely validated and accepted model and therefore kinematic outputs at extreme ROMs are likely to be trusted by the clinical community. However, this is not the case for SCM, and until further testing is carried out, kinematic measurement out with the limits of normal gait should be interpreted with caution.

From these results it can be concluded that the SCM kinematic output is comparable to the current clinical gold standard. PiG results exhibit some well evidenced anomalies such as high variability in internal/external rotation at the hip and knee, and kinematic crosstalk between knee flexion and ab/adduction. These should be taken into account when comparing the outputs. There are some consistent differences between the two models and repeated measures on individuals and groups should use one method or the other and the data compared to normal data collected with that model. It is not possible from these results to determine which model is most accurate, as it was not possible to compare PiG and SCM to a true measure of segment movement. This would require a form of bone embedded measurement which is out-with the scope of this study.

Chapter 5: Reliability of Strathclyde Cluster Model

Chapter 5

5 Reliability of Strathclyde Cluster Model

5.1 Introduction

Reliability is defined as "the extent to which measurements are consistent or free from variation" (McGinley et al., 2009). Cappozzo (1984) also stated that any protocol for movement analysis will only prove useful if it displays adequate reliability. Repeated measures are often used in movement analysis to determine change after a treatment intervention and therefore knowledge of error magnitude and reliability can minimise the risk of over interpreting small changes as meaningful (McGinley et al., 2009). However, quantifying reliability in motion analysis can be complex, as there are a number of sources of experimental variation to be considered. Further, natural variation exists between individuals and also within individuals between trials. However, this intrinsic variation should not be misinterpreted as extrinsic experimental variation such as that from errors in marker placement or anatomical landmark identification between sessions or assessors (McGinley et al., 2009; Schwartz et al., 2004). Figure 5.1 describes the different sources of error which contribute to the overall variability that may be observed between subjects or trials.

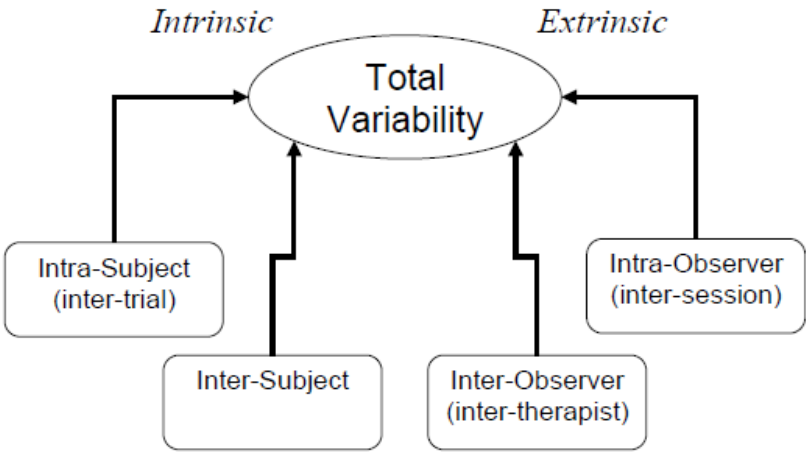


Figure 5.1 Contributions of different sources of error in motion analysis data (Schwartz et al., 2004)

Gorton et al., (2009) states that measurement variability can come from 3 primary sources: examiner, measurement system and subject. It was stated that the overall variability was defined as the sum of the variances from each independent source. They aimed to evaluate sources and magnitudes of variability in kinematic measurements on one subject over 12 motion analysis laboratories. One male subject was examined at 12 sites within a 3 month period. Examiners had a range of experience of 6 months to 21 years, with an average of 5 years' experience in GA. Results demonstrated a range of up to 28.3° for hip internal/external rotation between examiners with an average difference over all parameters of 14.8°. It was determined that more than 75% of overall variance could not be attributed to motion capture systems and the most likely result of variability between examiners was due to differences in marker placement.

Monaghan et al., (2007) tested intra-assessor reliability using a CODA motion analysis system and protocol (Figure 5.2).

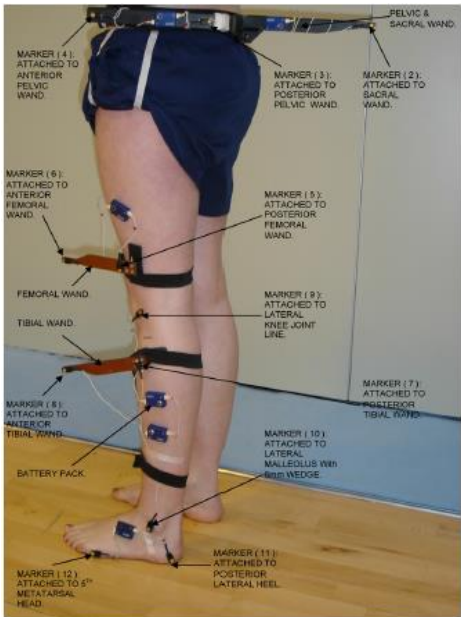


Figure 5.2 CODA gait analysis marker model (Monaghan et al., 2007)

Results demonstrated that spatiotemporal parameters (STPs) exhibited the least test-retest variability and kinematic parameters were more variable than kinetic. For kinematic and kinetic parameters, intra-assessor variability decreased with increasing number of trials. Table 5.1 details some potential sources of variability in test-retest experimental procedures which were identified.

Table 5.1 Potential sources of variability in test-retest experimental procedures. Adapted from Monaghan et al., (2007)

Variable	**Examples**
Subject	Natural gait variation
	Different footwear
	In response to lab setting (short runway)
	Real difference due to pathological change
System	Calibration
	Number of cameras and resolutions
	Relative skin/marker movement error
	Precision of computation algorithms
Assessor	Marker placement
	Identification of anatomical landmarks
	Anthropometric measurement
	Data processing

A review article by McGinley et al., (2009) stated that across 7 studies which employed a number of different protocols, the lowest obtained reliability indices (<

0.6) for within assessor reliability were reported for pelvic tilt, knee coronal plane and hip, knee and foot transverse plane movements. Sagittal plane reliability results were typically higher than 0.8, whereas coronal plane results were typically higher than 0.7 and transverse plane typically lower than 0.7. Three of 6 studies which investigated inter-assessor and intra-assessor error found single assessors to be more repeatable than multiple assessors. It was stated that the degree of acceptable measurement variation is directly related to the intended application and errors of 2° or less are likely to be considered acceptable. Errors of 2 to 5° are likely to be regarded as reasonable but should be interpreted with caution and errors in excess of 5° should cause concern and may be large enough to mislead clinical interpretation.

Stief et al., (2013) aimed to determine the reliability of two protocols, PiG and a custom protocol (MA) which used medial markers to eliminate reliance on thigh and tibial wand markers. Intersession reliability was examined on 10 healthy volunteers during 2 sessions separated by at least 3 days and within 2 weeks. Results showed that MA demonstrated higher reliability for coronal and transverse plane knee and hip angles than PiG. It was suggested that use of MA reduced crosstalk between knee flexion and abduction and error in marker placement was lower for MA than PiG. This confirms that high intersession and inter-assessor variability reported for PiG (McGinley et al., 2009) is likely to be caused by placement of thigh and tibial wand markers.

These results are supported by Kadaba et al., (1989) who investigated intra-subject repeatability with normal, adult subjects. Repeatability was assessed within a test day and between test days using GA and a standard HH style protocol. GA was performed on 40 normal adult subjects 3 times on 3 different test days at least 1 week

apart with a single, well trained assessor applying the markers each time. Results demonstrated that misalignment of thigh and tibial wand markers introduced a constant offset in coronal and transverse plane angles, however there was excellent within and between repeatability for sagittal plane angles.

It may be suggested from these studies that PiG and other conventional gait protocols display low intra-assessor and inter session reliability, particularly for movements out with the sagittal plane. Therefore, the aim of this investigation was to compare the reliability of PiG with SCM between and within assessors.

5.1.1 Methods

This study was approved by the departmental ethics committee at the Department of Biomedical Engineering, University of Strathclyde. One able-bodied participant (age – 23 yrs, mass – 63kg) and 6 assessors of varying levels of experience in GA (range – 6 months to 10 years, mean – 3 years) volunteered to take part in the study which took place in the Biomechanics lab in the Department of Biomedical Engineering, University of Strathclyde. All assessors attended a familiarisation session where procedures and marker placement for each model were outlined.

The same Vicon camera system described in section 3.8.1 was used and prior to each session the system was calibrated using the methods described in chapter 2. The participant wore tight fitting cycling shorts and a sports bra to allow palpation of anatomical landmarks and application of skin surface markers. In order to minimise inter session variability, the same participant was used for all sessions and the same clothing and shoes were worn for each session. Further, walking trials were performed on a treadmill which was set at a fixed speed of 1.19 m/s. On day 1 of the study, each assessor applied the PiG marker set to the participant and 20 seconds of

walking data were captured after a 2 minute familiarisation period. The assessor with most experience using PiG returned again in the same day and applied the marker set again in order to test intra-assessor variability. On day 2 of the study each assessor applied the SCM marker set to the participant and calibrated them using the calibration pointer described in section 3.2.2. The same process was then used to capture 20 seconds of walking data using SCM. The assessor who had the most experience using SCM also returned on the same day and applied the marker set again to assess intra-assessor reliability for SCM.

All marker trajectories were filtered using a 2^{nd} order Butterworth filter with a cut off frequency of 10hz and any further processing was carried out using the respective models. Kinematic data were normalised to 100% of the gait cycle (appendix 2) and coefficient of multiple correlation (CMC) was calculated for each kinematic output. A paired t-test assuming unequal variances ($\alpha = 0.05$) was used to determine any significant difference between CMC values between models.

5.1.2 Results

Similar trends were seen for the left and right sides, so only right side data is presented. Table 5.2 shows mean and SD inter-assessor CMC values for each kinematic parameter and each model. For the ankle, only 5 assessors were compared due to large systematic errors in data from one assessor.

Table 5.2 Mean (SD) inter-assessor (n = 6) CMC results for each right kinematic output and each model for 6 assessors (**only 5 assessors compared). *significant difference (α = 0.05)

Joint Rotation	PiG CMC Mean (SD)	SCM CMC Mean (SD)	P Value
Hip			
Flex/Extension	0.98 (0.01)	0.99 (0.004)	0.052
Ab/Adduction	0.96 (0.02)	0.93 (0.05)	**0.046***
Int/Ext Rotation	0.94 (0.02)	0.53 (0.24)	**<0.001***
Knee			
Flex/Extension	0.97 (0.02)	0.99 (0.01)	**0.047***
Ab/Adduction	0.73 (0.21)	0.85 (0.10)	0.06
Int/Ext Rotation	0.89 (0.07)	0.84 (0.08)	0.13
Ankle			
Plantar/Dorsiflexion**	0.74 (0.26)	0.76 (0.19)	0.91

For both models the majority of outputs demonstrated good reliability. Reliability can be said to be good if the CMC value is greater than 0.75, moderate if it is between 0.75 and 0.5 and poor if it is less than 0.5 (Collins et al., 2009). Good reliability was noted for all outputs from PiG except knee ab/adduction and ankle dorsi/plantar flexion. For SCM good reliability was reported for all outputs except hip internal/external rotation.

For the hip, CMC values were lowest for internal/external rotation and highest for flexion. SCM exhibited moderate reliability for hip internal/external rotation which resulted in a significant difference between PiG and SCM CMC values ($P<0.001$), with PiG exhibiting excellent reliability. A significant difference was also seen in CMC values for hip ab/adduction ($P=0.046$), with PiG exhibiting slightly higher reliability than SCM; however, both values were still excellent. There was no significant difference for hip flexion ($P=0.052$), with both models demonstrating excellent reliability.

For the knee, there was no significant difference in ab/adduction ($P=0.06$) or internal/external rotation ($P=0.13$) CMC values between models. However, there was a significant difference in flexion ($P=0.047$), with SCM indicating slightly

higher reliability than PiG. For the ankle, both models demonstrated moderate to good reliability and there was no significant difference in the CMC values (P=0.91).

Figure 5.3 provides a visual interpretation of variability between models. The mean ± 2 SD is presented for each kinematic output and each model. Variability was similar for both models although SCM appeared to be more variable for hip adduction than PiG. However, PiG appeared more variable for hip internal/external rotation than SCM. For both models, variability was higher for internal/external rotation than ab/adduction and variability was generally lowest for flexion.

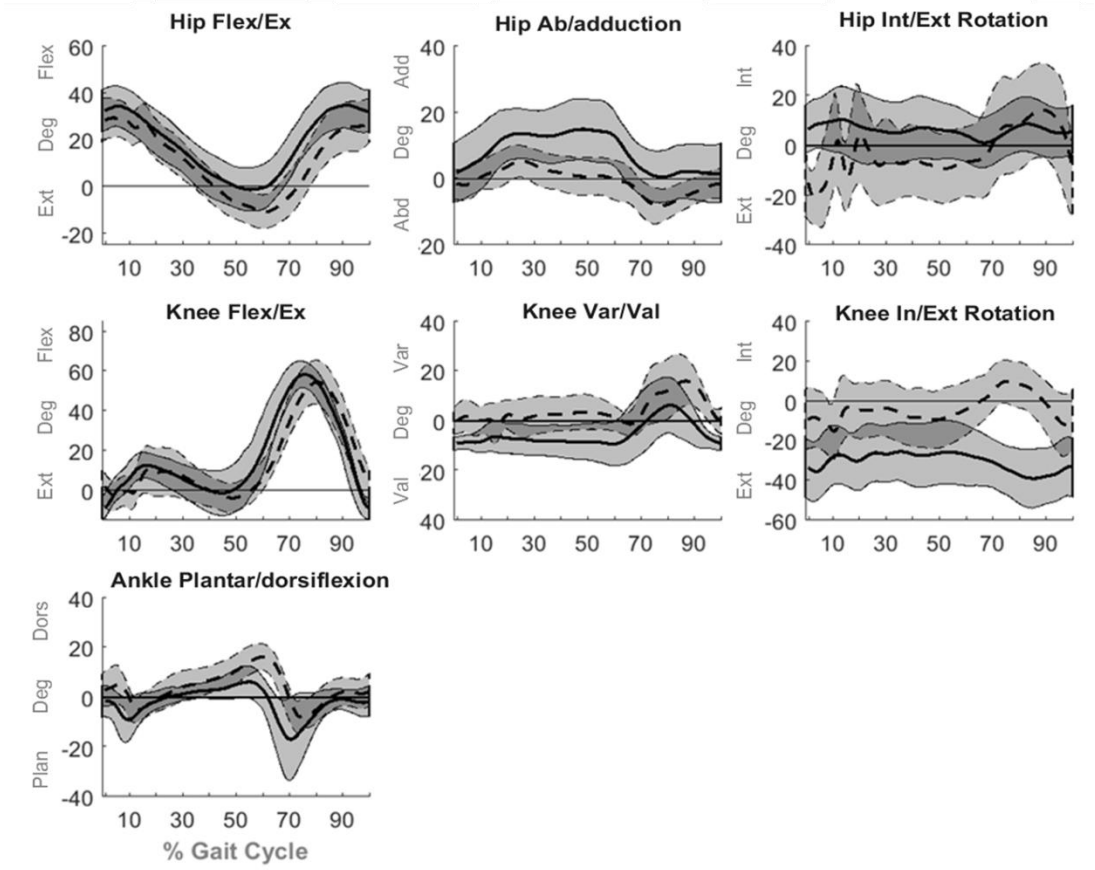


Figure 5.3 Mean kinematic variables of the participant's right leg for PiG (dashed) and SCM (solid) for 6 assessors. Shaded grey areas represent mean ± 2SD

Table 5.3 shows intra-assessor CMC values for PiG and SCM. CMC values were similar although slightly higher than those obtained for inter-assessor trials, most notably in hip internal/external rotation for SCM. Despite high values for all other

outputs, there is still moderate correlation between sessions for hip internal/external rotation for SCM. For hip flexion, PiG and SCM reported the same CMC values, whereas for adduction SCM reported a higher CMC. For the knee, PiG and SCM again reported the same values for flexion, however SCM reported higher values for abduction and internal/external rotation. The largest difference in CMC values between models was for knee abduction. For the ankle SCM reported a slightly higher CMC result than PiG, although both models exhibited excellent intra-assessor reliability. Figure 5.4 provides a visual representation of variability between models for intra-assessor trials.

Table 5.3 Mean intra-assessor CMC results for each kinematic output and each model for 2 trials

Joint Rotation	**PiG CMC**	**SCM CMC**
Hip		
Flex/Extension	0.99	0.99
Ab/Adduction	0.95	0.98
Int/Ext Rotation	0.89	0.59
Knee		
Flex/Extension	0.99	0.99
Ab/Adduction	0.89	0.98
Int/Ext Rotation	0.91	0.97
Ankle		
Plantar/Dorsiflexion	0.98	0.99

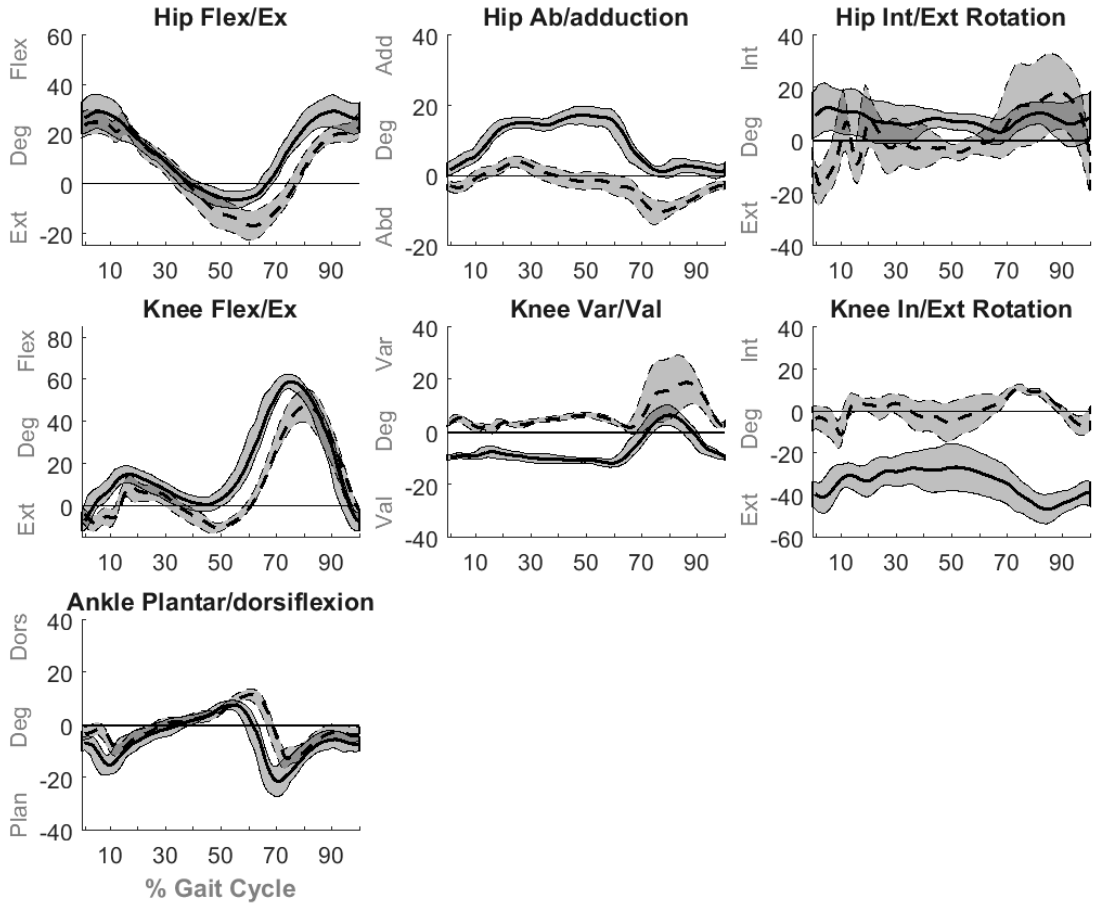


Figure 5.4 Mean kinematic variables of the participant's right leg for PiG (dashed) and SCM (solid) for 1 assessor over 2 sessions. Shaded grey areas represent mean ± 2SD

Variability for both models was similar, although PiG hip internal/external rotation variability was higher than SCM. Further, variability in knee abduction during swing was higher for PiG than SCM. Apart from PiG ab/adduction in swing, variability was low for both models for hip, knee and ankle flexion and hip and knee ab/adduction. Variability for both models was larger for internal/external rotation than ab/adduction or flexion.

5.1.3 Discussion

Generally, inter and intra-assessor reliability was high for both models. A notable exception is hip internal/external rotation for SCM, which exhibited moderate inter and intra-assessor reliability. Previous studies have noted lower reliability scores for

internal/external rotation than flexion/extension and ab/adduction (Collins et al., 2009; Kadaba et al., 1989). However, results reported in these studies are not as low as the result reported here for SCM.

This could be because CMC analysis is dependent on the range to which is applied. A high CMC result is more likely if the range of the data is large than if the range is small. For some joint rotations, SCM measured less excursion than PiG, therefore reducing the range of the data. This may have impacted on the CMC results and caused some values to be lower for SCM than PiG. Further, CMC analysis was only carried out on healthy individuals, which reduces the variability of data and therefore makes it harder to obtain a high CMC result, as all data are within a small confidence interval.

Collins et al., (2009) investigated inter-subject reliability within sessions and between sessions of a cluster based protocol (6DoF) and a modified HH protocol. Within session results demonstrated no significant difference in the sagittal plane for the hip; however, there was a significant difference in the coronal and transverse planes with HH displaying higher CMC values. For the knee there were significant differences in all planes with 6DoF displaying higher CMC values than HH. For the ankle, there was a significant difference in sagittal and coronal planes with 6DoF exhibiting higher CMC values and a significant difference in the transverse plane with HH exhibiting higher CMC values. All within session CMC values were high with the lowest value being 0.86 for ankle coronal rotations from the HH model. There were some similarities between data from Collins et al., (2009) and the current study. For both studies there was no significant difference in hip sagittal plane CMC values. For the knee, both studies reported a significant difference in flexion, with

the cluster model demonstrating higher CMC values. In the current study, knee abduction CMC values were also higher than PiG values, which is similar to Collins et al., (2009) however this difference was not significant. Both models in this study reported lower ankle plantar flexion CMC values than Collins et al., (2009). This may be due to differences in the way the foot and ankle complex is modelled. In the current study, the foot is modelled in PiG as a single vector, therefore the only reliable measurement which can be made is ankle plantar/dorsi flexion. However, Collins et al., (2009) report coronal and transverse plane rotations for the ankle, suggesting that it is modelled in such a way that allows these rotations to be estimated. This could account for the higher CMC values reported for the ankle in Collins et al., (2009). Further, the 6DoF model uses a functional method for estimating HJC location which may account for the higher CMC values for hip internal/external rotation reported for 6DoF in comparison to SCM.

Kadaba et al., (1989) assessed intra subject repeatability on 40 normal adults within and between days using a standard HH style marker set. GA was performed 3 times on 3 different test days, at least 1 week apart with a single, well trained operator performing all marker applications. For within day testing, CMC values for all kinematic outputs were high, with lowest values being reported for hip, knee and ankle internal/external rotation. Highest CMC values were reported for hip, knee and ankle flexion with all values being greater than 0.9. For hip and knee ab/adduction CMC values were also both greater than 0.9. Results from Kadaba et al., (1989) are similar to results from the HH marker set reported in Collins et al., (2009). This is likely to be because this marker set is a well-established model and has been modified and adapted over a number of years. As a result of this, higher

CMC values were generally reported for both HH protocols in Kadaba et al., (1989) and Collins et al., (2009) than SCM in this study. However, both studies used a single assessor, which is likely to result in higher CMC values than testing multiple assessors, as was shown in the current study.

Gorton et al., (2009) investigated the inter-assessor reliability of an HH style marker set at 12 different labs with a single subject over a period of 3 months. The mean was calculated for each kinematic output and the SD was used to estimate overall variability. Highest variability between sessions was seen in hip internal/external rotation (-5.0 ± 7.3) and lowest variability was seen in hip adduction. These results support CMC values for SCM in the current study, where repeatability was low for hip internal/external rotation. However, repeatability was not highest for hip adduction in the current study. The highest repeatability was measured for knee flexion by SCM, with a CMC value of 0.99. Gorton et al., (2009) reported a mean value of 20° for knee flexion, with an SD of 4.8, which was in the mid-range of all SD values measured. It can therefore be suggested that the results from Gorton et al., (2009) do not agree with the results from this study. However, this is unsurprising as a different marker model and method of analysis was used.

Monaghan et al., (2007) investigated intra-assessor reliability of a bespoke active marker setup using a CODA motion analysis system. Intra class correlation (ICC) was used to determine intra-assessor reliability of kinematic data on 2 separate days. Correlation was generally lower than results seen in this study. The highest correlation overall was reported for the ankle coronal plane (ICC = 0.97) and the lowest was reported for the hip sagittal and hip coronal planes (ICC = 0.56). Knee sagittal and transverse planes exhibited similar reliability (ICC = ~0.79), and there

was no report of knee coronal plane. Ankle sagittal plane results demonstrated high reliability (ICC = 0.94), which was similar to results obtained in this study. However, knee and hip reliability results for both models were higher for sagittal and transverse planes in the current study. These results could be explained by the use of a bespoke model by Monaghan et al., (2007) and could also depend on the level of expertise of the assessor. The experience of the assessor is not mentioned and therefore it could be that they had less experience with the model than the assessors in the current study. Few studies have investigated the effect of assessor experience on reliability of gait data (McGinley et al., 2009). However, palpation and location of anatomical landmarks and correct application of markers requires skill and an understanding of the biomechanical model and if the assessor is not experienced then this could have a significant effect on reliability results.

McGinley et al., (2009) conducted a review of gait data reliability studies. It was determined that comparison between studies of intra-assessor reliability was limited due to a number of different gait protocols and subject groups. Reliability varied across a number of studies although in general was highest for sagittal plane rotations and lowest for transverse plane rotations. The lowest CMC value was reported for knee internal/external rotation (CMC = 0.34); however this study was conducted on children and children's gait is inherently more variable than adult's gait (Stolze et al., 1998). The lowest value reported for healthy adults was for hip internal/external rotation (CMC = 0.41). These results generally agree with intra-assessor results obtained in the current study in that, for SCM the highest CMC values were observed for flexion and lowest were observed for internal/external rotation. For PiG, the hip also followed this pattern; however the knee did not, with internal/external rotation

displaying higher reliability than abduction. This could be due to the known issues with crosstalk occurring between flexion and ab/adduction in PiG (Ferrari et al., 2008).

CMC was used in this study to determine levels of agreement between and within assessors. Although previous studies have highlighted some issues with the use of CMC (Leardini et al., 2007), it is still considered useful as it expresses the ratio between true variability and error variability (Collins et al., 2009). However, plots of the mean for all assessors with confidence bands were provided in order to address some of the issues with CMC. This allows a visual assessment of levels of agreement between assessors and showed that levels of variability between assessors were similar for both models. One notable exception is that SCM exhibits higher variability in hip adduction than PiG. This is also reflected in the CMC values, as there was a significant difference in CMC values between models with PiG displaying higher reliability. However, the significant difference for hip internal/external rotation is not reflected in the variability of SCM data. It may therefore be suggested that 1 or 2 hip internal/external rotation outputs from SCM were not correlated, but didn't deviate far from the mean, thus resulting in a low CMC value and small confidence bands.

Further limitations of this study include the short time between each assessor applying each model as six assessors applied each model on each day. McGinley et al., (2009) stated that intervals between sessions should be far enough apart to minimise fatigue or memory bias but short enough to avoid genuine changes in measurements. Further, although artificially short intervals are easiest to achieve (within a day), these may leave visible signs of marker placement. This could have

been a possibility for PiG and may have contributed to the high CMC values reported. However, this is unlikely to be an issue for SCM as accuracy of kinematic output does not rely on placement of clusters, rather calibration of the participant, which is achieved using a pointer, therefore no visible marks would be left between sessions. This may also account for some of the lower CMC values reported for SCM in comparison to PiG. Additionally, fatigue may also cause true variations in gait, especially if all trials are being carried out in one day. In order to limit fatigue in this study, the participant walked for 2 minutes on the treadmill to allow familiarisation, and 20 seconds of data was then captured. Therefore, for each session only 2 minutes 20 seconds of walking was performed and as the participant was a young healthy individual, it is unlikely that fatigue would have contributed to any variation seen between assessors.

In general, intra-assessor reliability was higher than inter-assessor reliability. This has been reported previously (McGinley et al., 2009) and is unsurprising as intra-assessor reliability was evaluated using assessors with the most experience with each model whereas inter-assessor reliability was evaluated using 6 assessors with varying degrees of experience. One result which was consistent between inter and intra-assessor testing was hip internal/external rotation for SCM. Both inter and intra-assessor results reported poor reliability for this output. Generally, internal/external rotation is less reliable than flexion/extension or ab/adduction so this may be the reason for the low CMC value. Further, inter-assessor results were obtained with only one trial per assessor and intra-assessor results were obtained with only two trials. It may be possible to obtain better reliability results if more trials were obtained for both inter and intra-assessor testing.

In conclusion, SCM reliability was comparable to PiG for all joint rotations except hip internal/external rotation. This is likely to be due to the fact that PiG is a well-established model and has been developed over a number of years. SCM is a new model and 4 out of 6 assessors used it for the first time on the day of testing, whereas all assessors had some experience with PiG. The fact that such high CMC values were still obtained for SCM is promising and indicates that the reliability of the model is likely to be acceptable for clinical use.

Chapter 6: Development of Bespoke Feedback Scenarios

Chapter 6

6 Development of Bespoke Visual Feedback Scenarios

6.1 Introduction

This chapter introduces the concept of feedback and the importance of feedback in rehabilitation. The use of virtual reality to provide feedback to patients is discussed and a literature review of virtual reality in rehabilitation is presented. The development of bespoke visualisations to augment the rehabilitation process for patients who have undergone total knee arthroplasty surgery is also discussed along with patient acceptability of the visualisations.

The feedback of information to patients in the clinical environment is a key aspect in achieving a desirable outcome following a treatment intervention. Jones et al., (2011) state that "feedback on the performance of the action is essential for functional recovery". Generally, there are two main types of feedback: knowledge of results (KR) which would indicate task success and may aid in error correction for subsequent trials, and knowledge of performance (KP) which provides information about the nature of the movement pattern (Winstein, 1991). KR has been the focus of the majority of work on feedback and learning as it is easier to obtain, manipulate and quantify (Swinnen, 1996; Winstein, 1991). However, Jones et al., (2011) suggest that KP can inform patients about the control and coordination of their movements and may help to highlight any compensatory movements. Further, in a review by Subramanian et al., (2010) it is suggested that KP may lead to greater improvements in motor performance and quality. These studies would therefore suggest that KP is the best type of feedback to provide to patients during

rehabilitation and therapy, as KP relies on giving patients real-time information about their movement.

This type of feedback can be given in a number of different formats, from verbal reinforcement to full-scale virtual reality (VR) environments. Jones et al., (2011) propose that providing the patient with augmented visual feedback of their movement could have a positive effect on rehabilitation outcomes. Further, there is extensive evidence indicating the importance of providing patients with information regarding their performance in order to learn a new task or complete an exercise (Levinger et al., 2016; Swinnen, 1996; Todorov et al., 1997).

Mirrors can be used to provide real-time feedback; however the movement is perceived backwards and therefore may be confusing. Video recording can also be used although this does not provide real-time feedback. Further, both these solutions limit visualisation to one plane and rely on patients looking at a true image of themselves, rather than a digital representation, which may be distracting or distressing. Therefore, a more appropriate way to provide this type of feedback is through the use of motion capture technology, described in chapter 2. A limited number of studies have been conducted into the use of visualisation feedback in the clinical environment; however, visualisation feedback and virtual reality are routinely used in the elite sporting community to improve performance (Bideau et al., 2010; Gorman, 2012; Randell et al., 2011). The use of digital video and motion analysis technology in sport to assess athletes' movement is commonplace and is often referred to as performance analysis (Bampouras et al., 2012; Groom et al., 2011; McGarry, 2009). Performance analysis employs a number of feedback techniques which would easily be transferrable to the clinical environment if the

technology were available. Advantages of the use of performance analysis in sport include coaches and researchers having access to athletes concerns, intentions, sensations, emotions, expectations and interpretations. It also allows changes in these dimensions to be monitored (Sève et al., 2013).

Gorman, (2012) investigated the effect of augmented feedback in the form of verbal cues and delayed video feedback on elite swimmers' entry angle into the water. After several months of training with augmented feedback, results demonstrated that the swimmers' entry angle and distance had improved.

Randell et al., (2011) investigated the effects of real-time feedback with pro-rugby players. Two groups of players completed a squat jump training programme; one group received real-time feedback on a screen during training and the other group received no feedback during training. Results suggested that the feedback group performed better than the non-feedback group. It was suggested real-time feedback resulted in greater consistency of effort and performance throughout the programme.

While the majority of feedback in sport is video based, Bideau et al., (2010) stress the advantages of VR over video playback. It was stated that "VR lets the researchers or coaches systematically control and tune all factors affecting the player's judgement". In other words, the amount of information being fed back can be controlled. For example, if using an avatar of movement, there is the option to display the whole avatar, or just the segments of interest. Visual cues for the movements required can be added, removed or altered depending on the desired outcome and biomechanical information can be displayed if required. De Freitas and Oliver, (2006) suggest that "games and simulations can become an effective way of accelerating learning outcomes". The importance of an immersive environment is

also stressed, as this would easily allow the development of outcome specific games or simulations.

These studies suggest that the positive effects of augmented feedback have been proven in healthy and active individuals and that VR can be an invaluable tool when training a new task or refining a specific movement.

Holden, (2002) stated that "VR is a simulation of a real-world environment that is generated through computer software and is experienced by the user through a human-machine interface". It is suggested that for non-neurologically impaired individuals, the principle of learning through augmented feedback from VR is well established (Holden, 2002). The main advantages of VR include safety, i.e. there are fewer negative consequences to failure, automated documentation and less time needed to change or setup equipment between tasks. VR also facilitates 'purposeful gaming' which can make practice and adherence to a regime more fun and therefore increase compliance. Rose et al., (2000) stated that "virtual reality embodies many characteristics of an ideal training environment." It is also suggested that VR results in increased motivation and allows complete control over the learning environments and pattern of feedback.

Rose et al., (2000) aimed to investigate the use of VR to train motor skills and whether the learned task would transfer to real-life situations. Two groups practiced a hand-steadiness task whereby a wand was moved along a piece of wire and the objective was to prevent the wand from touching the wire. One group trained in real-life and one group trained using VR only. Results suggested that not only did learning in VR result in adequate performance in real-life, the VR group outperformed the real-life group when it came to performing the real-life task,

suggesting that the use of VR can enhance the performance of motor tasks in real life situations.

It has been established that the provision of augmented feedback can enhance motor learning in non-neurologically impaired individuals (Holden, 2002). Todorov et al., (1997) tested the use of VR in learning a complex table tennis shot in healthy adults. One group learned the shot with no feedback and the other used VR in which a virtual teacher's paddle was superimposed over their own, showing how the shot should be performed. The results suggested that augmented feedback resulted in performance benefits in comparison to the controls, although the difference was not significant.

Further studies have also suggested that feedback using VR can improve motor learning in stroke survivors and neurologically impaired individuals. MacDonald et al., (2009) developed a software tool aimed at augmenting stroke rehabilitation by allowing visualisation of movement from sensors attached to the body (Figure 6.1).

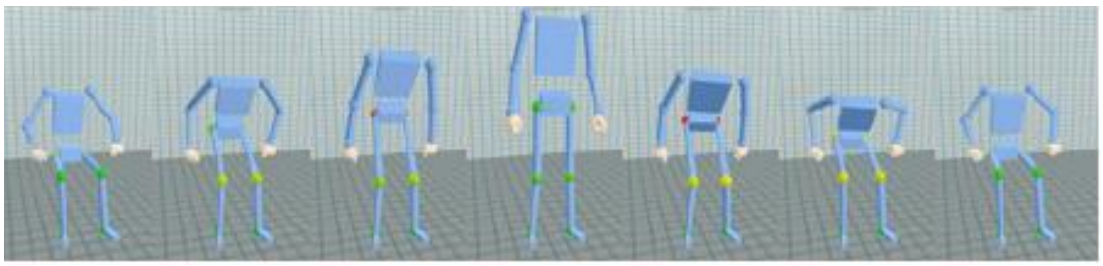


Figure 6.1 Example of visualisation software demonstrating a patient performing a sit to stand task. A "traffic light" system was used to indicate levels of stress on the joints. Green indicates low levels of stress, yellow indicates medium stress and red indicates significant levels of stress (MacDonald et al., 2009)

A key finding of this work was improved communication between the patient and clinician and improved understanding of biomechanical data. Further, older adults felt capable of participating in the discussion of problems with their mobility.

Holden (1999) conducted a pilot study which assessed the effects of training upper limb motor tasks in stroke survivors using VR. Three main research questions were posed. Can subjects with hemiplegia improve in a virtual task following virtual practice? Does learning that occurs in VR transfer to a similar real-life task? Does learning in VR transfer to related but untrained tasks, or to functional activities not specifically trained? Two patients were recruited, one 3.5 years post stroke who reported no functional use of the affected arm and another, 1.5 years post stroke who reported limited functional use of the affected arm. The virtual task involved posting a letter into a mailbox at several different positions. Each subject received treatment sessions for 1-2 hours once or twice a week, resulting in 16 total sessions. Results showed that virtual performance improved over the 16 sessions as both participants could only complete easy levels to begin with but were both able to complete the hardest level by the end of their sessions. Further, both subjects exhibited positive transfer of their improvement to similar real-life tasks. It cannot be concluded that feedback from the VR was the cause of the improvement, as no controls were implemented. It could therefore be suggested that the patients may have also improved if they had performed the tasks in real-life. However, evidence suggests that goal-setting and purposeful gaming have a positive effect on functional outcome and may increase compliance (Chao et al., 2015; Webster and Celik, 2014; Wingham et al., 2015).

Merians et al., (2002) implemented the use of a VR system to augment rehabilitation training aimed to improve the motor function of the hand in 3 patients in the chronic phase after stroke. The VR had a performance meter which indicated how close participants were to the movement required. Goal targets could also be adjusted for

each participant depending on their ability. Participants had sessions lasting 20-25 minutes 4 times a day and completed a total of approximately 30 sessions. Following the VR sessions all participants showed improvement in strength, ROM and speed of completing a task.

These studies indicate that use of VR can result in improved motor learning with stroke patients and positive transfer to real-life functional tasks. However, the majority of studies implemented relatively intense training schedules which may not be easily transferred to realistic clinical environments. It is suggested that current use of VR to provide feedback in everyday clinical situations is prevented by the complexity and cost of the systems which are needed to provide feedback. Further, there is a lack of studies investigating the use of VR in realistic clinical situations which is also likely to limit its clinical use. However, (Carse et al., 2013a) implemented the use of VR for patient and clinician feedback in a more realistic clinical environment. Visualisation software was used to aid ankle-foot orthosis (AFO) tuning for stroke patients. A randomised controlled trial tested the outcome of feedback on the walking velocity of patients who were prescribed and fitted with an AFO. Intervention patients received AFO fitting and tuning using the visualisation software whereas control patients received AFO fitting and tuning by standard observation. Results showed that intervention patients displayed an immediate increase in walking velocity compared to controls. It was suggested that the biomechanical information augmented the AFO tuning process, resulting in a more appropriate orthosis for the patient. Further, the feedback of information during the process may have also contributed to the positive result.

These studies suggest that the use of VR in rehabilitation may be a promising avenue for improving patient experience and functional outcome. With the introduction of more mainstream VR systems designed for video gaming, the possibility of delivering this type of rehabilitation is more accessible than ever. Two of the most popular commercial VR gaming systems are the Microsoft Kinect™ (Microsoft UK, Berkshire) and the Nintendo Wii™ (Nintendo of America, Redmond, WA, USA). The Kinect incorporates infra-red light and a video camera to create a 3D map and is capable of automatically detecting anatomical landmarks on the body such as joint centres without the need for skin surface markers (Clark et al., 2012). The Wii uses a handheld controller instrumented with a gyroscope and accelerometer. Games are controlled through this interface but the Wii is in incapable of measuring gross whole body movements. An extension to the Wii is the Wii Fit™ which is a balance board similar in concept to a forceplate. Games can be controlled through balance exercises and feedback can be given on weight distribution and balance control (Fung et al., 2012). A number of studies have investigated the effectiveness of using the Wii and Kinect as a cheaper alternative to motion capture in order to provide a VR environment for rehabilitation.

Webster and Celik (2014) conducted a review into Kinect-based early care and stroke rehabilitation systems. It was reported that Kinect applications have been developed for fall detection and gait assessments to reduce fall events. Sufficient accuracy was reported for gait events such as stride and stance time and limb angular velocities, but occlusion caused by adjacent objects prevented the use of Kinect sensors for a home-based fall detection system. It was also suggested that Kinect may overcome stroke patients' unwillingness to perform rehabilitation exercises by

providing an engaging activity. Kinect based gaming had a positive effect on users' emotional wellbeing as well as functional outcome. Results from a number of Kinect based systems showed significant improvements in emotional state, physical function, pain, visual performance skills, reaction time and hand-eye coordination.

Further, a number of studies have also reported improvements in outcomes after use of the Wii and so-called exergames during rehabilitation for a number of patient populations (Chao et al., 2015; Park et al., 2014). Fung et al., (2012) investigated the effect of using the Wii Fit as part of rehabilitation for TKA patients. Twenty seven patients were in the intervention group and 23 were in the control group. Outcome measures included active ROM, 2 minute walk test, numeric pain rating scale, lower extremity function scale, activity specific balance confidence scale and length of outpatient rehabilitation. No significant differences were found between groups although the intervention group did show better function in the 2 minute walk test and higher scores in the numeric pain rating scale, activity specific balance confidence scale and lower extremity function scale. It was concluded that Wii Fit has the potential for use as an adjunct to physiotherapy treatment for TKA patients. In this study, both the lower extremity function scale and activity specific balance confidence scale are self-report questionnaires. The higher scores reported for the intervention group could therefore be due to the positive psychological effect of the feedback which the equipment provided. This study also demonstrates that the use of force and balance data can be useful in TKA rehabilitation and should be considered in any VR feedback system.

Baltaci et al., (2012) aimed to determine the acceptability of the Wii Fit compared to conventional rehabilitation as a therapy tool for patients with ACL reconstruction.

Thirty male participants with a mean age of 28 were recruited and half were placed in a conventional rehabilitation group and half were placed in a Wii rehabilitation group. The Wii group completed 4 sports based games as well as balance board exercises for 15 minutes at each rehabilitation session. Outcome measures included coordination, proprioception, response time and dynamic balance. No significant differences were found between groups for any outcome measures at any stage of rehabilitation. However, this may be due to the fact that all participants were young and active. This may have caused increased drive and compliance to programmes in both groups in order to return to sporting activities and hence negated the positive effects of VR feedback in young, motivated individuals.

Lozano-Quilis et al., (2014) developed a Kinect based VR system for use in motor rehabilitation with multiple sclerosis patients. An experimental group of 11 patients spent 15 minutes performing virtual exercises after their standard rehabilitation session. Virtual exercises included touching a virtual object, moving a virtual object and stepping over virtual objects to reach a target. Outcome measures were static and single leg balance, dynamic balance and a suitability evaluation questionnaire. Significant improvements were seen in the experimental group for static, dynamic and single leg balance.

Chao et al., (2015) conducted a review to summarise the impact of using exergames in older adults. It was suggested that many older adults may not adhere to standard rehabilitation programmes due to lack of enjoyment and therefore Wii exergames could be a feasible way to prevent low compliance due to boredom. It was also stated that, in general, the use of exergames has improved outcomes in TKA patients, acute older adults' rehabilitation and patients with Parkinson's disease. Further,

Wingham et al., (2015) explored the perceptions of people and their caregivers following stroke in using the Wii for rehabilitation. People with arm weakness after stroke who used a Wii at home reported wanting to use it due to its perceived effectiveness, instantaneous feedback and their interest in most games.

From these studies it is clear there is proof of concept evidence that the use of VR, feedback and exergames has a positive effect on a number of different patient populations. The use of VR creates a stimulating and motivating environment which can help encourage patients to complete exercises.

However, the use of the Wii does not enable measurement of gross movement which may lead to patients performing exercises with the incorrect mechanism of action, but still in theory completing the exercise.

The Kinect claims to be capable of tracking gross body movements (Galna et al., 2014) and a number of studies have investigated its use for measuring movement in a clinical setting. Clark et al., (2012) aimed to assess the concurrent validity of the anatomical landmarks as measured by the Kinect compared with a Vicon MX system during 3 standing postural control tests (single leg standing balance, forward reach and lateral reach). Twenty young, injury free participants were recruited. Results demonstrated excellent concurrent validity for anatomical landmark displacement and trunk angle data when compared to Vicon. However, errors in Kinect data increased with increasing movement amplitudes. Although the results demonstrated excellent concurrent validity, all movements were very small and were carried out by healthy individuals. Further, Kinect was unable to cope with large movement amplitudes which is likely to limit its use for a wide variety of clinical training and assessment of functional movements.

Baldewijns et al., (2014) evaluated the results of an approach to measure step length and step time using Kinect compared to the GAITRite walkway. Two healthy adults were investigated. Kinect provided a comparable output for step time and step length. However, some issues were encountered with participant detection. A number of trials had to be discarded due to participants not being detected or only parts of participants being detected. From this study it was suggested that, on its own, Kinect is not an acceptable tool for clinical measures of gait.

Galna et al., (2014) aimed to assess the accuracy of the Kinect to measure functional and clinically relevant movements in people with Parkinson's disease. Data were captured concurrently for a group of patients with Parkinson's disease using the Kinect and Vicon. Healthy adults were also tested. Movements tested included standing still, reaching forwards, reaching sideways, stepping forwards, stepping sideways, walking on the spot, hand clasping, finger tapping, foot tapping, leg agility, sit to stand and hand pronation. The main outcome measures were mean ROM during the task and the timing of the task. Results demonstrated that Kinect was able to accurately measure the timing of movements. However, there were significant differences in ROM kinematics and joint positions and the error increased with increasing ROM. Further, it was suggested that estimating movement while seated may introduce error when using Kinect as the legs of the chair may mistakenly be identified as the participant. It was concluded that Kinect can accurately measure timing and gross spatial characteristics of clinically relevant movements; however, its current use in the clinic is limited by the inability to accurately measure movements of large magnitude and the lack of user friendly software. Results of this study also suggest that Kinect would be unsuitable for use

with any patient who requires a walking or standing aid as this would interfere with participant detection. This is supported by Auvinet et al., (2015) who determined that heel strike detection with Kinect is difficult because of confusion between the ground and the participant.

Xu and McGorry (2015) aimed to examine the accuracy of Kinect joint centre coordinates during various static postures when compared to 3D motion capture. Twenty healthy participants were recruited and the following postures were tested: upright standing, left lateral bending, right lateral bending, trunk flexion, left foot raise, right foot raise, stoop and squat. For lower extremity joint centres Kinect resulted in an inaccuracy level of over 100mm and in some cases up to 452mm.

From these studies it is clear that Kinect is not currently an acceptable method for delivering accurate and reliable joint kinematic data in the clinical setting. While it may be suitable for implementing VR during motion controlled exergames, kinematic parameters and gait events should be interpreted with caution depending on the type and magnitude of the movement. Further, Kinect is not currently equipped with an appropriate user interface and therefore a level of technical understanding is likely to be required for correct use in the clinic.

From the literature it is clear that the use of VR has positive outcomes for a number of patient groups. However, current systems are not capable of providing an appropriate standard of clinical measurement of kinematics on which to base training or outcome assessments. Therefore, there is a need for a system which can provide VR, feedback, accurate and reliable movement data and an appropriate user interface for clinical use. Thus, the aim of this part of the study was to develop a tool which can provide feedback of movement in the clinical environment. Further, as few

studies have reported qualitative feedback from patients regarding augmented feedback, the acceptability of such a tool was also assessed.

6.2 Methods

Evidence from the previous section suggests that a virtual representation of movement, such as an avatar, displayed in a virtual environment is likely to be an acceptable method of providing feedback regarding movement (MacDonald et al., 2009). Therefore, for this study, an avatar which represented real-time lower limb movement and was displayed in the virtual environment within D-Flow was developed. Figure 6.2 represents D-Flow's virtual environment and the X (red), Y (green) and Z (blue) axes which comprise the GRF. During all exercises patients faced in the negative Z direction.

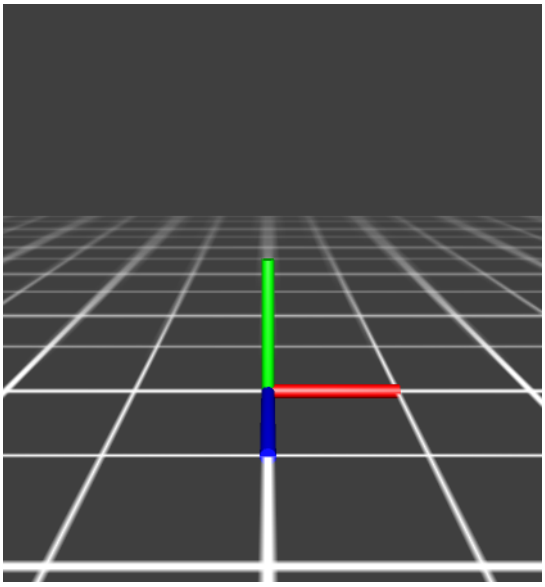


Figure 6.2 Virtual environment within D-Flow. Axes represent the global reference frame. X – red, Y – green, Z – blue

When developing their software tool, MacDonald et al., (2009) gave patient focus groups a number of options regarding which type of avatar they found most favourable. Patients were given the option of an avatar which closely resembled a human, one which was a skeleton and one which was a simplified 'segment' human.

Qualitative data suggested patients did not respond favourably to an avatar which closely resembled a human. They felt it was distressing to see a close representation of themselves struggling with movement tasks. The avatar which resembled a skeleton was also rejected as it was less aesthetically pleasing. Further, the developers were concerned about users assuming that real skeletal measurements were being made. As described in chapter 3, segments are modelled as rigid bodies but, without embedded bone pins, accurate measurement of bone movement is not possible. The third option of a simplified segment human was more widely accepted by developers and patients. It was close enough to a real human to mimic life-like movements, but not so close as to cause distress. Therefore, a simplified segment human avatar was developed for this tool (Figure 6.3). MacDonald et al., (2009) also discussed how patients were comfortable viewing only the segments of interest when performing movement tasks. Therefore, an avatar with legs, a trunk and head was deemed acceptable for this tool. As discussed in chapter 3, the biomechanical model for this study encompasses the lower limbs only. Therefore, the addition of a trunk and head to the avatar are for aesthetic purposes only and are driven by movement of the pelvis.

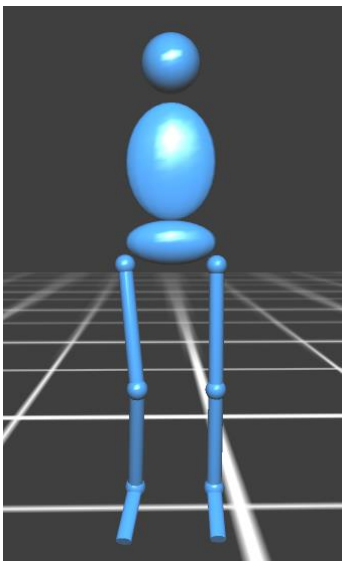


Figure 6.3 Avatar of human movement. Only the lower limbs are driven by the biomechanical model, the trunk and head are extensions of the pelvis for aesthetic purposes

The SCM was used to drive the avatar by linking segments between joint centres. A spherical object was placed at each joint centre. Equation 6.1 and Equation 6.2 were used to place a cylinder in between two joint centres to create a segment graphical object. Equation 6.1 positions the segment in the XZ plane and Equation 6.2 positions the segment in the XY plane, where Δx, Δy and Δz are the x, y and z distances between joint centres, respectively and Δxz is the xz distance between joint centres. Rotating the segment about its Y axis by rotY and about its Z axis by rotZ brings the distal end of the segment in contact with the distal joint centre, while the other end is attached to the proximal joint centre. The X axis of the segment is the long axis and no rotation about this axis is required as the segment graphical object is homogeneous in shape and colour. Segment object axes are defined by D-Flow and are not the same as segment ARF axes. The code used to create the avatar is presented in appendix 1.

$$rotY = \tan^{-1}\left(-\Delta z/\Delta x\right) \times 180/\pi$$

Equation 6.1

$$rotZ = \tan^{-1}\left(\Delta y/\Delta xz\right) \times 180/\pi$$

Equation 6.2

Bespoke feedback scenarios were then developed for three exercises (appendix 1) which are commonly performed during knee rehabilitation: single and double leg step-up, sit to stand (STS) and a weight transfer exercise. Further, a user-friendly graphical user interface (GUI) was developed to allow operators with limited technical expertise to calibrate a patient and run a session incorporating the 3 exercises. The GUI is made up of a number of tabs which allow the operator to control the software (Figure 6.4). Together, these components formed the basis of a visual feedback tool which was termed the Lower Limb Visualisation Tool (LLVT).

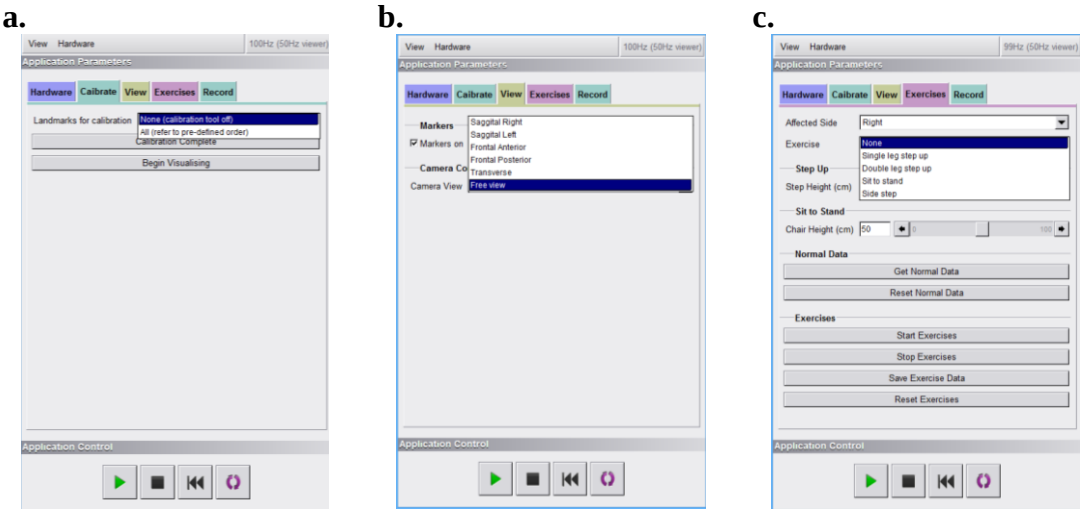


Figure 6.4 Graphical user interface for the LLVT **a.** Calibration tab **b.** View tab **c.** Exercises tab

Within the GUI, the calibration tab (Figure 6.4a) allows the calibration tool to be off or on by selecting 'None' or 'All', respectively. To calibrate the patient, the anatomical landmarks of the SCM model are recorded using a pointer, as described in chapter 3. Selecting 'All', allows the patient to be calibrated and once all

landmarks have been identified, 'Calibration Complete', saves the A to T^F matrices for each segment (chapter 3) and 'Begin Visualising' creates the biomechanical model and displays the avatar.

The view tab (Figure 6.4b) controls what is shown and allows visualisation of cluster and anatomical landmark markers which can be used to check if all cluster markers are visible and also that virtual markers have been reconstructed correctly. Further, multiple camera views are available, with the default being a free view, allowing the user complete control over the camera position. Other views in the form of sagittal, coronal and transverse planes are also available, allowing the user to return to a pre-determined view at any time.

The exercises tab (Figure 6.4c) allows setup and control of all 3 feedback scenarios. First, the user selects which side is the affected side and then which exercise is to be performed by the patient. For the step up and STS exercises, step and chair height can be set to allow visualisation of the step and chair in the virtual environment to reflect that of the real environment. Prior to performing exercises, a small number of repetitions are performed to determine 'normal' thresholds of movement for each patient. 'Get Normal Data', allows this information to be stored and 'Reset Normal Data' allows it to be cleared between patients. Once the non-affected repetitions have been completed, the patient is ready to begin exercises. Exercise feedback scenarios start automatically upon completion of the non-affected repetitions; however 'Start Exercises' can also be used if the exercises were not being completed immediately following the unaffected repetitions. Similarly, each feedback scenario has a timer and stops automatically when the timer counts down to zero. 'Stop Exercises' allows exercises to be stopped earlier than the pre-determined time.

Further, once a patient has completed the exercises, 'Reset Exercises' clears the virtual environment of the current feedback scenario and allows a new feedback scenario to be loaded. Patient data from each exercise can also be saved using the 'Save Exercise Data' button. Specific methods for how each feedback scenario was developed are described in the following sections.

6.2.1 Bespoke Feedback Scenarios

All code which was used to implement feedback scenarios is presented in appendix 1. Similar methods were employed for all feedback scenarios; therefore, methods for the step up exercise will be discussed in detail to describe how feedback mechanisms were developed and any other additional methods employed in the STS or weight transfer exercises will then be reported. The aim of the step-up exercise was to increase ROM and strengthen the quadriceps on the affected side. During a step-up TKA patients may circumduct their leg and 'hitch' the hip to avoid having to flex the knee and normally activate the quadriceps muscle while still achieving the step up movement. Therefore, the aim of this feedback scenario was to provide patients with guidance on completing a step up whilst minimising lateral deviation of the affected limb and achieving as much flexion of the affected knee as possible. In order to mimic the real environment as much as possible, a step object was placed in front of the patient, in line with the centre of their pelvis and at a height suitable for them which was specified in the GUI.

In order to minimise lateral deviation of the affected leg during the step up, red, amber and green bands were introduced which the patient had to aim to keep their leg within (Figure 6.5).

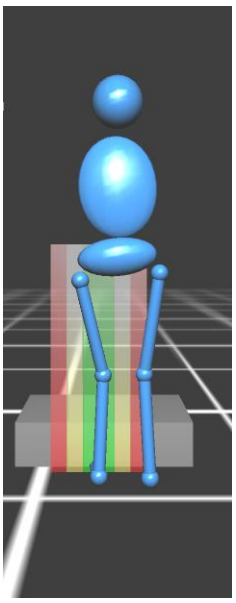


Figure 6.5 Red amber and green bands help prevent patients from deviating laterally while performing a step up

In order to tailor the feedback scenario to each patient, 5 repetitions were performed on the non-affected side before the feedback scenario was constructed. During those 5 repetitions, maximum lateral deviation of the non-affected knee was measured and used to define the size and position of the green, amber and red bands. The size of the bands was determined by 2 thresholds. Threshold 1 (T1) was the maximum lateral deviation of the KJC on the non-affected side and threshold 2 (T2) was the maximum lateral deviation multiplied by 2. The width of the green band was T1 multiplied by 2, indicating a 95% confidence interval of normal movement on the non-affected side. The width of the amber and red bands was T2-T1. The bands were positioned in the YZ plane, with the centre of the green band directly in front of the affected KJC, using Equation 6.3 through Equation 6.5.

$$green = KJCx$$

Equation 6.3

$$amber = KJCx + T1 + \left({}^{T2 - T1}/_{2}\right)$$

Equation 6.4

$$red = KJCx + T2 + \left({}^{T2 - T1}/_{2}\right)$$

Equation 6.5

As well as encouraging patients to perform the step up with limited lateral deviation, the aim for this feedback scenario was also to encourage as large a ROM as possible. To achieve this, the maximum knee flexion angle was measured during the 5 repitions performed on the non-affected side. This was then used as a goal which the patient aimed to achieve with their affected side. The maximum knee flexion angle achieved for each repetition on the affected side was also shown to allow patients to see how close they were to the aim (Figure 6.6).

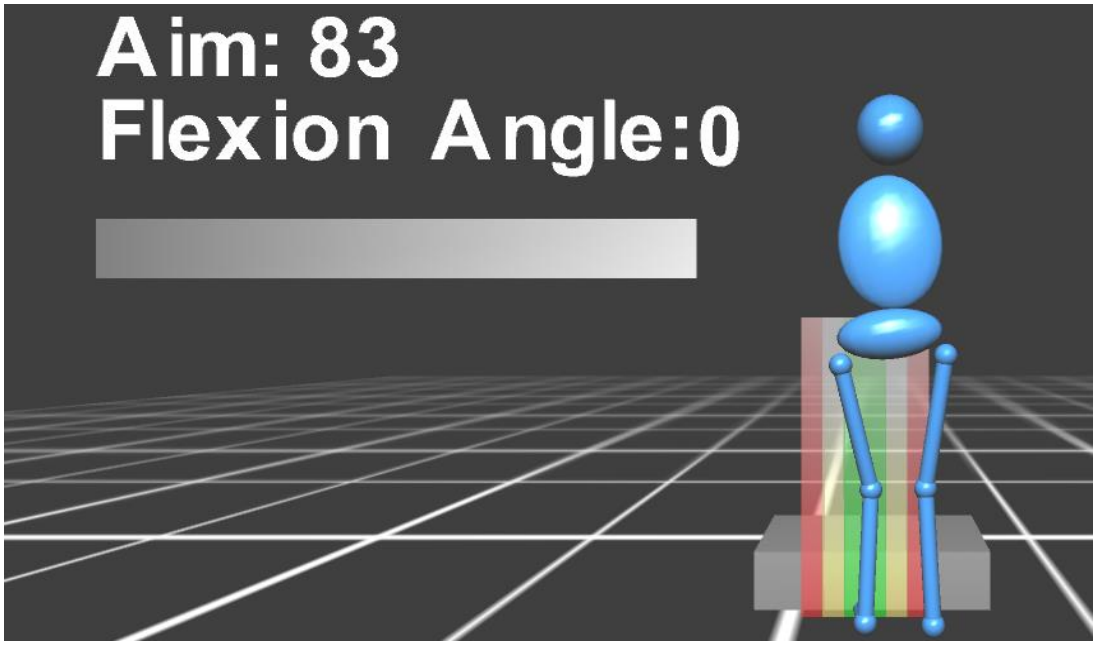


Figure 6.6 The maximum knee flexion achieved on the non-affected side is displayed as an aim. The maximum knee flexion achieved on the affected side during each rep is shown for comparison

The maximum flexion angle was calculated in conjunction with the number of repetitions performed using a simple function (Figure 6.7).

```
function getMaxAngle(p1, p2, jointAngle)
    kinematicData = kinematicData or {}
    maxValue = maxValue or {}

    if p1 > p2 then
        start = 1
    else start = 0
    end
    if p1 < p2 then
        finish = 1
    else finish = 0
    end

    if start == 1 then
        table.insert(kinematicData, jointAngle)
    end

    if #kinematicData ~= 0 and finish == 1 then
        maxJointAngle = math.max(unpack(kinematicData))
        table.insert(maxValue, maxJointAngle)
        kinematicData = {}
    end

    nrReps = #maxValue

return nrReps, maxValue
end
```

Figure 6.7 Function used to determine number of reps and maximum joint angle for each rep

In order to determine when a repetition was taking place, the height of the AJC (Figure 6.7; p1) and height of the step (Figure 6.7; p2) were used such that if the AJC was higher than the step, a repetition was taking place. While a repetition was taking place, knee flexion angle was stored in a table and as soon as a repetition finished the maximum value recorded during that repetition was also stored in a different table. This process was repeated for each repetition and therefore the number of entries in the maximum value table was also the number of repetitions completed.

A large sphere in the virtual environment also gave patients feedback on whether the ROM they achieved for each repetition was ‘good’, ‘medium’ or ‘bad’ by turning green, yellow or red, respectively (Figure 6.8).

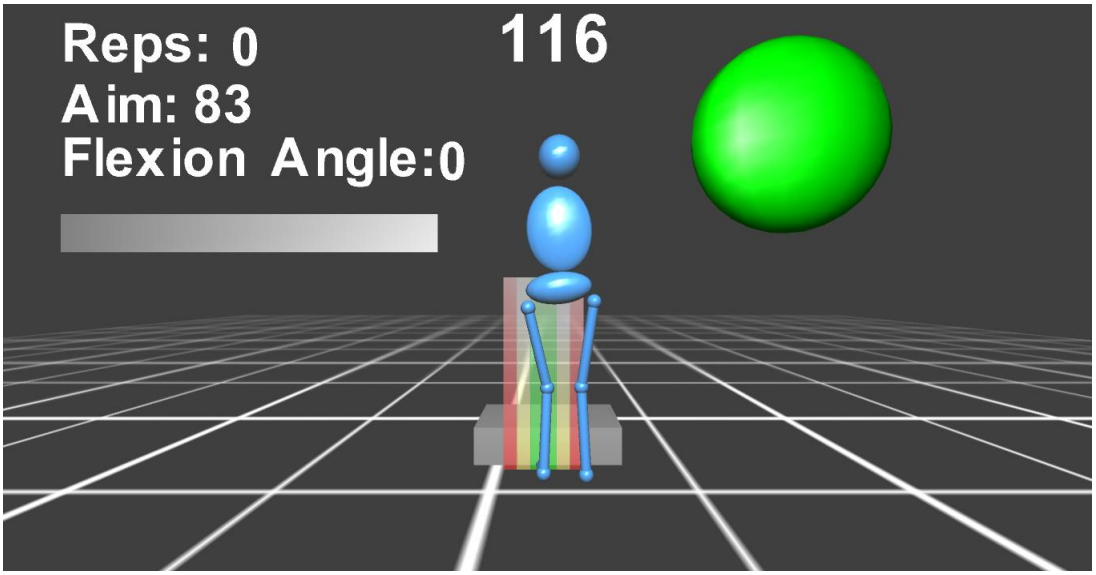


Figure 6.8 Whole feedback scenario for step up exercise. The sphere on the right turned green, yellow or red depending on whether the patient performed a good, medium or bad rep, respectively

Repetitions were determined as good, medium or bad depending on how close the patient was to the aim. Table 6.1 details the thresholds which determined whether a repetition was good, medium or bad.

Table 6.1 Joint angle thresholds and corresponding rep qualities

Threshold	**Rep Quality**
Maximum joint angle ≥ aim – 5°	Good
Maximum joint angle < aim-5° and ≥ aim -10°	Medium
Maximum joint angle ≤ aim -10°	Bad

After discussion with the physiotherapist in charge of the patients' rehabilitation, it was determined that these thresholds would be an appropriate guide as most patients should be able to achieve a majority of good repetitions towards the end of their rehabilitation. In order to increase the motivational effect of the feedback, every time a good repetition was performed a section of a progress bar filled up with green. The aim was to perform at least 10 good repetitions, at which point the progress bar would be fully green and the patient would receive a firework and a motivational message would be displayed. The progress bar would then be reset (Figure 6.9).

Further, upon completion of the exercises, patients were provided with the percentage of good, medium and bad repetitions they achieved (Figure 6.10).

Figure 6.9 Firework and motivational message which appeared when 10 good reps were completed

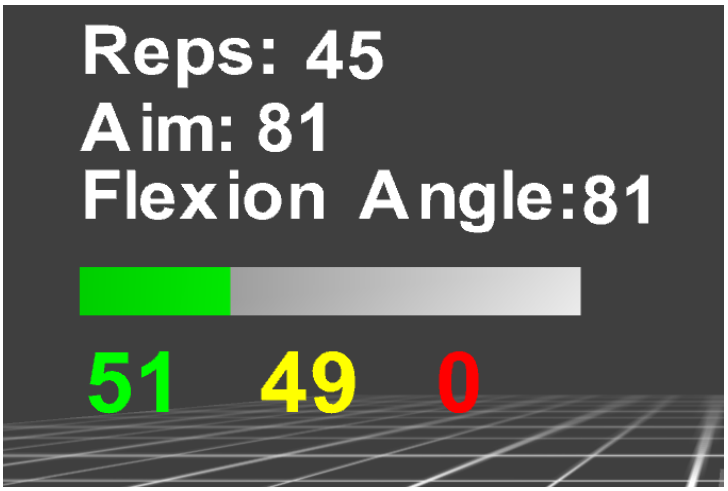


Figure 6.10 Patients were provided with the percentages of good, medium and bad reps they achieved in green, amber and red text, respectively

For the STS exercise, the aim was also to increase ROM and strengthen the quadriceps on the affected side. The focus during this exercise was to prevent excessive medial movement of the knee as, during a STS, patients may exhibit excessive medial movement due to quadriceps weakness. Further, patients should aim to achieve as large a ROM as possible during the movement. Figure 6.11 is an

example of a STS feedback scenario which employs many of the same methods as the step up feedback scenario.

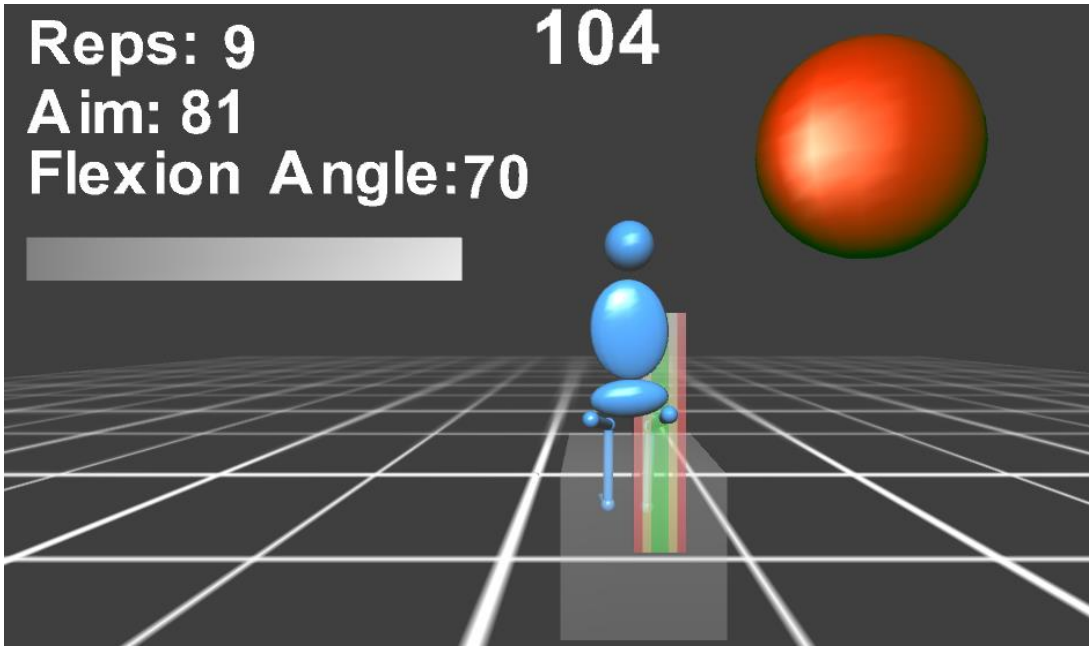


Figure 6.11 Example of sit to stand feedback scenario

Instead of a step, a box was placed in the virtual environment to represent a chair, at the height specified in the GUI. A small degree of transparency was introduced to the box to allow patients to have full view of the lower limbs whilst retaining a first person view. Red, amber and green bands were used to prevent excessive medial deviation of the knee, which were created and positioned using the same methods as those for the step up. However, medial deviation of the non-affected knee was used to set the thresholds instead of lateral deviation. A similar function to that used for the step up (Figure 6.7) was used to count the number of repetitions and calculate the maximum angle on the affected side. However in this case, p1 was the initial height of the pelvis as the patient was standing in front of the box, and p2 was the live height of the pelvis. A repetition was taking place if p1 < p2-5cm and not taking place if p1 > p2-5cm. This also ensured the patient stood up fully between each

repetition in order for it to count. All other aspects of the STS feedback scenario were created and executed using the same methods as described for the step up.

Figure 6.12 shows an example of the feedback scenario for the weight transfer task. The aim of this exercise was to transfer as much weight as possible on to the affected side by moving the hips sideways. This would ideally build quadriceps strength and allow the patients to experience accepting weight on their affected knee. Since this exercise relied on transference of weight, use of a forceplate would have been the optimal method for ensuring proper execution. However, as the clinic was not equipped with a forceplate, the methods devised were instead used to ensure proper execution.

Some similar methods were used for this feedback scenario; however there were a number of aspects which differed from the step up and STS feedback scenarios.

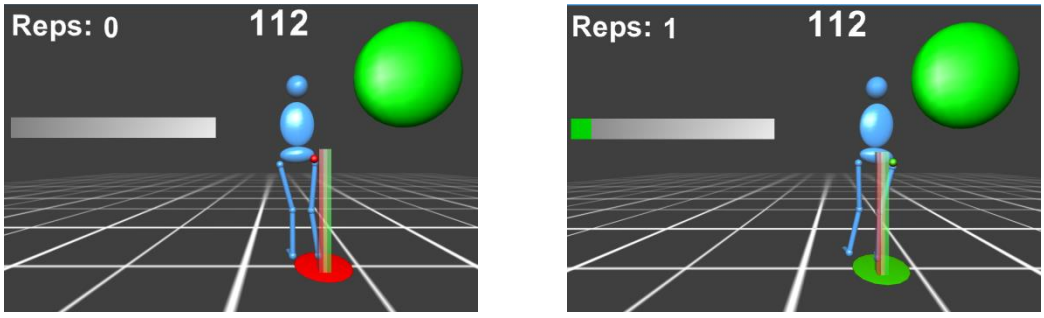


Figure 6.12 Example of weight transfer feedback scenario

Again, red, amber and green bands were used; however, instead of acting as a guide for lateral movement of the knee, they were used as a target. The aim was for the patient to move the ASIS on the affected side as far into the bands as possible, aiming to hit the green band with each repetition. The ASIS position was marked by a small red sphere (Figure 6.12), which reflected the y and x coordinates (height and mediolateral position, respectively) of the ASIS; however the sphere's z coordinate (anterior-posterior position) was positioned at ASISz+20cm. This moved the sphere

away from then avatar enough so that patients could see its position while performing the exercise.

The position and width of the bands was determined using the previously described methods, whereby the patient performed a small number of repetitions on the unaffected side to determine normal ROM. A repetition was defined using a similar function to the one described in Figure 6.7, where p1 was the live x coordinate (mediolateral position) of the affected ASIS and p2 was the initial x coordinate of the affected ASIS. Since knee flexion was not a goal for this exercise, an aim and flexion angle were not provided. Instead, the aim was to move the small sphere which marked the ASIS position as far into the bands as possible, aiming to get into the green band with each repetition. As the patient moved through the bands, the small sphere and the circle on the floor changed from red, to amber and then green, depending on which band the ASIS was currently within. Once the repetition was complete, the large sphere changed to green, amber or red depending on which colour band the patient reached. As with the step up and STS, a progress bar filled up with green each time a good repetition was performed. Video examples of each exercise are located in appendix 1.

6.2.2 Qualitative Feedback from Patients

In order to determine the acceptability of visual feedback with patients, an outcome questionnaire was developed (appendix 3) to determine how well patients responded to the feedback. The questionnaire aimed to determine how patients felt about wearing cluster markers and being calibrated, how they felt about seeing a virtual representation of themselves, if they understood the information which was displayed

on the screen and also whether they enjoyed using the tool or not. Table 6.2 details each statement on the questionnaire.

Table 6.2 Statement numbers and accompanying statements in the patient questionnaire

Statement Number	Statement
1	I found the cluster markers comfortable to wear
2	I found the calibration process (palpation of bony landmarks) comfortable
3	I felt comfortable seeing a virtual representation of myself on the screen
4	I was comfortable aiming for a virtual target
5	I understood the biomechanical information which was displayed on the screen
6	I found the biomechanical information useful in helping me to complete the exercise
7	I found the visualisations helped me understand how I was moving
8	I felt that I could discuss the information on the screen with my physiotherapist
9	I enjoyed using the visualisation tool as part of my rehabilitation

Patients were asked to rate how much they agreed or disagreed with each statement using a 5 point scale ranging from strongly agree to strongly disagree. All patients completed weekly rehabilitation sessions for 6 weeks in a class based, outpatient environment and received visual feedback for 3 of 9 exercises each week. Visualisation was provided on a TV monitor in front of the patient as they carried out exercises within the motion capture volume (Figure 6.13).

Figure 6.13 A patient performing a sit to stand exercise within the motion capture volume and with visual feedback on a screen in front of them

Other exercises also focused on increasing quadriceps strength and improving ROM at the hip and knee (appendix 6). Patients who experienced using the tool were asked to fill out a questionnaire on completion of their rehabilitation. Questionnaires were completed by patients with no input from the researcher, except to clarify the meaning of statements.

6.3 Results

Only qualitative feedback from patients will be presented in this chapter, quantitative results regarding the effectiveness of feedback in rehabilitation will be presented in chapter 7. Fourteen of 15 intervention patients returned completed questionnaires following use of the tool and Figure 6.14 shows the results of the questionnaire feedback.

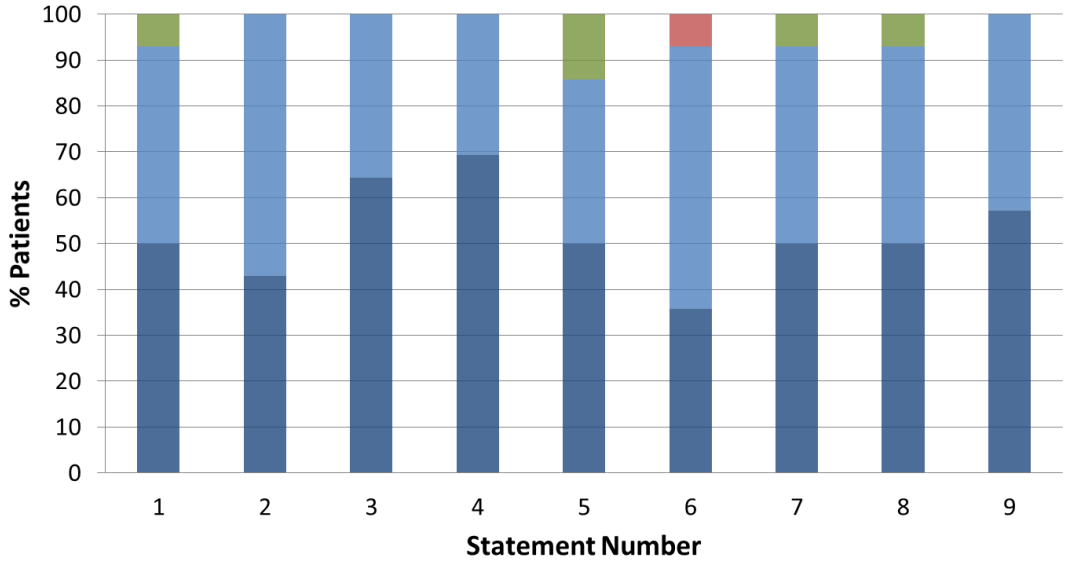


Figure 6.14 Results of patient feedback questionnaires. Dark blue: strongly agree, light blue: agree, green: neither agree nor disagree, red: disagree

The majority of patients either strongly agreed or agreed with all statements. Statement 4 showed the highest percentage of strongly agree responses (I was comfortable aiming for a virtual target) whereas statement 6 (I found the biomechanical information useful in helping me to complete the exercise) saw the lowest. Further, statement 6 was the only statement to receive a disagree response. Some patients neither agreed nor disagreed with statements 1, 5, 7 and 8, whereas the majority of patients strongly agreed with statements 3, 4 and 9. For statements 2 and 6, the majority of patients only agreed, as opposed to strongly agreed. However, in general the patients who experienced the visual feedback strongly approved of it. Patients were also asked to state anything they liked or did not like about the tool. Table 6.3 details some of the responses from patients regarding use of the tool.

Table 6.3 Responses from patients regarding use of the lower limb visualisation tool

Positive Responses	**Negative Responses**
"I would have liked a saved record of how I was doing"	"I would have liked a saved record of how I was doing"
"I didn't want the machine to beat me"	"I didn't want the machine to beat me"
"I could see how I was managing"	"I didn't really pay attention to the information"
"I can see my progress"	"It made me tired"
"Helpful – motivating"	
"I liked the fact you could watch your progress every week"	
"I liked seeing the improvement"	
"Found it helpful in helping me work towards correcting movements"	
"Helped to motivate and challenge me to do better"	
"A very helpful, self-motivating tool"	
"The information was like a traffic light"	
"I didn't get any green so I'll need to practice that one at home"	

Some responses could be interpreted as positive or negative; however the majority of responses were positive, with a number of patients saying they found it helpful and motivating. A number of patients stated they liked being able to see their progress and could see improvements week on week. One patient additionally stated that they would have liked to see a record of how they had done in previous weeks in order to better compare their weekly performance, indicating the value of the feedback and showing that it was motivating. Two negative comments were made where one patient stated that they did not pay attention to the biomechanical information and another said it made them tired, although this could have been due to them exercising more intensely which could be positive.

6.4 Discussion

The aim of this section was to develop a tool which could provide feedback and measurement of movement in the clinical environment and assess the feasibility of

such a tool with TKA rehabilitation patients in a realistic clinical environment. To assess how well the LLVT was accepted by patients, qualitative analysis of their experience was carried out using a questionnaire with a 5 point Linkert scale ranging from strongly agree to strongly disagree. For the vast majority of statements, patients either agreed or strongly agreed although there was a small percentage of responses which were 'neither agree nor disagree' and 'disagree'. More than 50% of patients strongly agreed with statements 3, (I felt comfortable seeing a virtual representation of myself on the screen) 4, (I was comfortable aiming for a virtual target) and 9, (I enjoyed using the visualisation tool as part of my rehabilitation) suggesting these were the most acceptable aspects of the tool. These statements were related to patients being comfortable with a virtual representation of themselves and whether they enjoyed using the tool. This would suggest that most patients don't have a problem with the avatar and can relate what is happening on the screen to their own movement. Further, it suggests the majority of patients enjoyed seeing their movement in this way. There is currently little evidence to support whether patients enjoy and can accept seeing their movement as an avatar. One previous study gathered patient feedback regarding visualisation with an avatar and the response was also positive (Loudon et al., 2012). However the majority of studies which investigate the use of visualisation and virtual reality are not yet at the stage to be tested with patients (Webster and Celik, 2014), therefore there is very little data on patient response to feedback of this type.

Less than 50% of patients strongly agreed with statements 2 (I found the calibration process (palpation of bony landmarks) comfortable) and 6, (I found the biomechanical information useful in helping me to complete the exercise) suggesting

these may be less strongly acceptable aspects of the tool. Statement 2 was regarding the calibration process and whether patients found it comfortable, so perhaps, in future, steps should be taken to improve the calibration process to make it more comfortable for patients. The lower percentage of 'strongly agree' responses for this statement could have been due to the prolonged periods of standing required for the calibration process, particularly at the baseline stage where patients may only be a few days post-op, or due to the palpation of bony landmarks which may have been uncomfortable, particularly on the affected knee. The introduction of functional calibration would eliminate uncomfortable palpation of landmarks, however it may not eliminate long periods of standing and may also require patients to perform ROM tasks which they may find even more uncomfortable than simply standing for a few minutes (Piazza et al., 2001). Further questioning would be required to determine which aspects of the calibration process patients found most uncomfortable.

The fact that less than 50% of patients strongly agreed with statement 6 (I found the biomechanical information useful in helping me to complete the exercise) may suggest that while patients enjoyed using the tool, they may not have necessarily used the information in the way it was intended. The largest proportion of 'neither agree nor disagree' was for statement 5 (I understood the biomechanical information which was displayed on the screen). This suggests that the information aspect of the tool is the part which is likely to need most improvement as a number of patients didn't agree with these statements as strongly as they had for others. This may be because they didn't understand the information, or because they didn't pay attention to it. Further questioning would be required to gauge the numbers of patients who paid proper attention to the information, and those who didn't. Statements 7 (I found

the visualisations helped me understand how I was moving) and 8 (I felt that I could discuss the information on the screen with my physiotherapist) also had a small proportion of patients who answered 'neither agree nor disagree' and only 50% of patients who strongly agreed. These results suggest that although patients could understand and relate to an avatar of their movement, it didn't automatically help them, in all cases, to recognise specific abnormalities in their movement patterns. One reason for this could be the small size of the avatar and screen used in the study. The screen used to provide the feedback was smaller than was ideal, and therefore the avatar had to be small in order to fit the whole feedback scenario in the screen space. This may have masked some of the more subtle movement abnormalities which may have been picked up if the avatar was larger and clearer. In future, feedback should be provided on as big a screen as possible, or more ideally, projected onto a clear wall in front of the patient.

A number of patients also felt less strongly about being able to discuss the information with their physiotherapist. There could be a number of reasons for this. First, some patients said they felt less strongly about being able to understand the information which may have been a reason they wouldn't want to discuss it with their physiotherapist. Second, there was only one physiotherapist for a class full of patients so it may have been likely that there wasn't time for every patient to discuss every exercise with the physiotherapist. In future, the tool may be better used in a one to one environment where the physiotherapist can dedicate all their attention to the patient to ensure they understand the information and can answer any questions the patient may have. Another possible solution could be the introduction of a 'virtual teacher'. A number of studies which investigated visual feedback provided

subjects with a virtual example of how the task should be carried out (Holden, 1999; Todorov et al., 1997). Todorov et al., (1997) concluded that expert movement presented in the same coordinate frame as the subject's movement should be provided to ensure maximum effectiveness of the visual feedback. Further, Holden (1999) reported an improvement in performance of upper limb function for 2 stroke patients after providing them with visual feedback which included a virtual teacher superimposed over the patient's own movement. Neither of these studies presented participants with any biomechanical information, which suggests that it may be more useful to provide a virtual teacher which demonstrated the appropriate ROM for each patient, instead of presenting the patients with an aim in the form of a number. This may reduce the likelihood that patients wouldn't understand the information and may also increase their willingness to pay attention to the screen, as they have something which they need to copy.

However, in general, the feedback was overwhelmingly positive. Many patients said they liked that they could see their progress and improvement and one patient said they would have liked a record of how well they were doing week to week. This suggests that patients want to remember what they achieved in previous weeks and therefore in future it may be worth providing them with the scores they achieved the week before so they can compare. Many patients said they found the tool motivating, although there was one patient who experienced the opposite effect and became distressed when they were unable to achieve amber or green in any of the exercises. This suggests it is important to find the balance between being motivating and being too challenging and also shows the importance of performing this type of rehabilitation in a therapeutic environment where these concerns can be addressed.

The idea of generating the exercise objectives based on what the patient could achieve on their good side was to individualise the goals of the therapy. Perhaps for some patients this is still too hard, particularly during the first and second sessions which are so soon after surgery. It may be that the method for generating patient specific aims should be revised and an option should be in place to make the exercises easier if there is a patient who appears to be struggling.

One patient said "I didn't get any green, I'll need to practice that one at home", suggesting that the effects of the feedback may have extended past those seen in the class. It is likely that if the patient had been performing the exercise on their own, with no feedback, they may not have realised that more practice was necessary. Another patient said the tool made them tired, which could be taken one of two ways: either the effect of having to concentrate on the screen and understand the information made them tired, or the challenging aspect of the exercises made them tired. However, the patient who wrote this also stated that they didn't really pay attention to, or understand the information. This makes it more likely that it was the challenging aspect of understanding the feedback which made them tired, in which case, this is a negative response. Another quote which could be taken positively or negatively was "I didn't want the machine to beat me". This may have been negative as it may suggest that the patient didn't like using it and didn't enjoy the challenging aspects of the feedback. However, this patient selected 'strongly agree' for statement 9 (I enjoyed using the visualisation tool as part of my rehabilitation) which therefore suggests that it motivated them to try harder and not that they didn't enjoy using the tool.

There are a limited number of studies which have used visual feedback during rehabilitation, and of these studies, none appear to have reported qualitative feedback from patients regarding their experience, and certainly not with the sample size of the current study. MacDonald et al., (2009) did collect qualitative feedback from focus groups during the development of their visualisation tool; however, feedback from patients during routine use has not been reported. One study (Noort et al., 2014) did assess the acceptability of 4 different types of feedback on healthy participants during gait training. Participants were asked to decrease their knee adduction moment or increase their hip internal rotation angle and were presented with 4 different types of visual feedback to help them achieve this. They were also asked to report on the intuitiveness of each type of feedback. The feedback ranged from a colour coded traffic light system, to a real-time graph of the parameter of interest. There was no significant difference between ratings for each type of feedback although the colour coded feedback was rated "most intuitive" when modifying knee adduction moment and the graph was rated "most intuitive" when modifying hip internal rotation angle. This suggests that acceptability of the feedback may be dependent on the parameter which is to be modified, although conclusions about this cannot be drawn from this study alone. Further, the current study did not investigate the effects of different types of feedback, although the positive response to a colour coded system seems to be present in both studies.

A limitation of this part of the study was that results were obtained using qualitative analysis which has been less widely accepted by the scientific community than quantitative analysis (Collingridge and Gantt, 2008). Quantitative analysis is most widely accepted in scientific research as it employs experimental methods to analyse

the causal relationship between two variables (Golafshani, 2003). Qualitative analysis is any kind of research that produces findings which were not arrived at by means of statistical procedures or other means of quantification (Golafshani, 2003) therefore, it may be highly subjective. For these reasons, it is often not employed in scientific methodology; however, in order to determine the experience of patients who used the tool, it was deemed that some qualitative analysis was necessary. The results of this qualitative analysis support the perceived suitability and benefit, from the user's point of view, of the visual feedback developed.

Chapter 5 discusses the effectiveness of the feedback using quantitative analysis techniques and it has been noted that a combination of qualitative and quantitative analysis can be used in conjunction to build and refine a theory (Shah and Corley, 2006). Therefore, while the qualitative analysis does not show the intervention was effective, it does show that the patients perceived it as worthwhile and it was suitably designed for them to be able to benefit from it. There was a lack of qualitative feedback from patients and users of augmented feedback tools in the reviewed literature. However, the patients' experience should not be ignored during development of these types of tools. The overall effectiveness of an augmented feedback tool, no matter how accurate or reliable it is, is very much dependent on the patient experience, because if patients don't like it, then this is likely to impact on any positive effects the tool may have.

In conclusion, patients were comfortable seeing a virtual representation of themselves which agrees with previous studies which used visual feedback in the form of an avatar (Loudon et al., 2012). Further, patients could accept wearing the cluster markers and having to go through the calibration process and they mostly

enjoyed using the tool as part of their rehabilitation. However, some improvements could be made regarding the biomechanical information which was displayed. In future, it may be helpful to convey the information in a clearer manner, perhaps using a virtual teacher, so that all patients can understand and follow it. The effectiveness of the intervention will be discussed in the following chapter.

Chapter 7: Effectiveness of Visual Feedback in TKA Rehabilitation

Chapter 7

7 Effectiveness of Visual Feedback in TKA Rehabilitation

7.1 Introduction

As discussed in chapter 1, patients who have undergone total knee arthroplasty (TKA) surgery are likely to have reduced functional outcome compared to healthy controls (Benedetti et al., 2003; Ouellet and Moffet, 2002). One aspect which could be contributing to this less than normal functional outcome is the nature of rehabilitation following surgery. Despite agreement that rehabilitation is necessary following TKA (Brander and Stulberg, 2006; Nickel and Botte, 1992) there has been limited investigation into the biomechanical effects of rehabilitation on gait and functional outcome in TKA (Callaghan and Oldham, 1995; Jevsevar et al., 1993; Lowe et al., 2007). Further, any studies which did assess improvement in function investigated active and passive knee ROM, SPT gait parameters and QoL using questionnaires regarding pain and function, as opposed to movement of the knee as expressed by gait kinematics (Codine et al., 2004; Frost et al., 2002; Kramer et al., 2003; Mockford and Beverland, 2004; Moffet et al., 2004).

A number of authors have suggested that visual feedback during rehabilitation may have a positive effect on functional outcome for stroke patients (Carse et al., 2013a; MacDonald et al., 2009; Merians et al., 2002). It has been hypothesised that biofeedback based rehabilitation for TKA patients may potentially enhance recovery as surgery frequently affects knee proprioception, which may impair sensorimotor function (Levinger et al., 2016; Pap et al., 2000). However, only 2 studies in the reviewed literature investigated the effects of rehabilitation with visual feedback on functional outcome following TKA (Levinger et al., 2016; Zeni et al., 2013).

Levinger et al., (2016) aimed to investigate the effectiveness of biofeedback using a Kinect and a Wii on patients who had undergone TKA surgery. The investigation was a case-study of one patient compared to 3 controls and all patients undertook a 6 week in-patient rehabilitation programme, with controls receiving rehabilitation with no feedback. The biofeedback software was designed to incorporate real-time visual feedback aimed at correcting limb alignment, movement pattern and weight distribution (Figure 7.1).

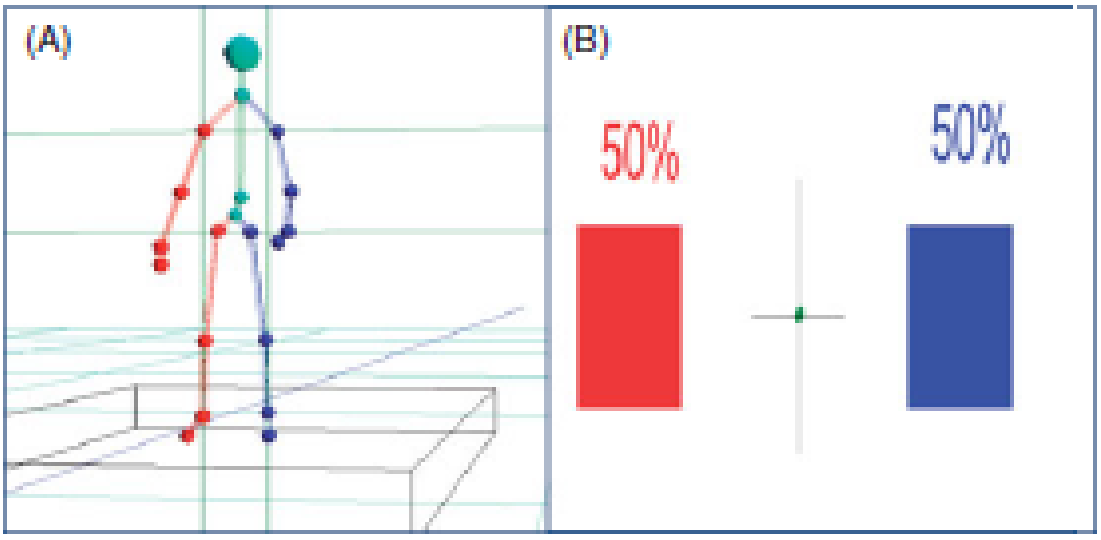


Figure 7.1 Visual display used to help patients distribute weight equally between their affected and unaffected sides (Levinger et al., 2016)

Both the controls and the biofeedback patient demonstrated improvements in pain, function and QoL; however, the biofeedback patient showed a better improvement in knee moment pattern, indicating the possible effectiveness of a biofeedback-augmented intervention. However, it was stated that the Kinect was very sensitive to glare and bystanders in the field and it was also noted that it was not capable of providing an accurate measurement of knee flexion, although this was justified by the fact that the knee flexion angle was being used purely for feedback, and not for functional assessment.

Zeni et al., (2013) developed an outpatient rehabilitation strategy to promote symmetrical weight bearing during functional exercises using visual feedback. The aim was to conduct a cohort study to test the feasibility and effectiveness of integrating the protocol into outpatient rehabilitation following TKA. Nine protocol patients were tested prior to surgery, at discharge from physical therapy and at 3 and 6 months post-op. A group of control patients who did not receive symmetry feedback were also tested at 6 months post-op. All patients received outpatient rehabilitation 2 to 4 weeks post-op, 2 to 3 times a week for 6 to 10 weeks, depending on their progress. Feedback on symmetry was provided using the Wii balance board during STS and wall slide tasks. Results suggested that the symmetry protocol was well tolerated by patients and produced functional outcomes which were equivalent or superior to controls. Further, it was reported that the symmetry protocol group had more symmetrical quadriceps strength than the control group and they had a sagittal knee moment pattern which was similar to normal controls, whereas the control group did not. However, it was not reported whether there was a significant difference between groups at the 6 month post-op testing. It was only stated that there was a clinically meaningful difference between pre-op and 6 months post-op for the protocol patients. Control patients were only tested at 6 month post-op and therefore it was not stated whether there was also a clinically meaningful difference for this group.

While these were the only two studies in the reviewed literature which investigated the effects of visual feedback on TKA rehabilitation, one further study investigated the effects of visual feedback in total hip arthroplasty (THA) rehabilitation (Pataky et al., 2009). The aim of this study was to investigate the effectiveness of a

biofeedback method based on information about foot pressure intensity to reduce percentage bodyweight loading following THA. Eleven patients were recruited and the PEDAR in shoe system was used to provide feedback to patients in the form of a graph with a threshold. The ground reaction vector (GRV) was compared at the beginning and end of the learning period, which lasted a few days, and at 3 stages of retention: 10 minutes, 1 day and 2 days. Results demonstrated no significant difference in the magnitude of the GRV between the beginning of the learning period and different retention points. It was concluded that even with visual feedback, suitably low GRVs could not be obtained at any retention tests.

While there are limited studies investigating the effects of visual feedback in orthopaedic rehabilitation, a number of studies have investigated the effects of visual feedback in gait training on healthy individuals, with a view to application in OA (Barrios et al., 2010; Noort et al., 2014; Wheeler et al., 2011). All studies reported positive changes in kinematic and/or kinetic gait parameters in response to visual feedback; however, they were all conducted on healthy participants and so do not give an indication of the effects of visual feedback during rehabilitation for TKA.

The majority of studies which have investigated gait following TKA do not focus on rehabilitation and have follow-up times much longer than 6 weeks. Ouellet and Moffet (2002) compared the gait of TKA patients and controls 2 months post-op. It was not stated if controls were age matched and whether the patients received any rehabilitation. However, some abnormalities in gait were detected for TKA patients. It was determined that at 2 months post-op, the replaced knee had a reduced ROM and TKA patients had a longer stance time in comparison to controls. Benedetti et al., (2003) evaluated 9 patients at 6, 12 and 24 months post-op, all of whom received

specific gait training once they could fully weight bear on the affected side, which was approximately 2 weeks after surgery for most patients. Some gait abnormalities were reported at all follow-up stages, mainly reduced knee flexion, particularly at mid stance, toe-off and during swing.

Smith et al., (2004) tested TKA patients pre-surgery and 12-18 months post-surgery. It was not mentioned if patients received any rehabilitation; however, post-surgery, TKA patients exhibited reduced knee flexion in loading and reduced maximum knee flexion in swing in comparison to controls. Further, Myles et al., (2006) tested TKA patients at 4 and 18-24 months post-op and determined that total knee joint excursion was reduced in comparison to controls at both follow up stages.

The reviewed articles suggest that the introduction of biofeedback during TKA rehabilitation may have a positive effect on gait and functional outcome and that currently the outcome from TKA is less than functionally ideal.

Therefore the aim of this part of the study was to determine if augmented rehabilitation in the form of visual feedback improved the functional outcome of patients who have undergone TKA and if so, if they were fully rehabilitated in terms of their gait.

7.2 Methods

This study was completed in partnership with NHS Ayrshire and Arran at the Musculoskeletal Centre, Biggart Hospital, Prestwick, Scotland. Ethical approval was granted by the NHS Ethics Committee, West of Scotland REC1, Western Infirmary, Glasgow.

The study population consisted of TKA replacement patients who were within their first few weeks of post-op recovery. Patients were provided with information about

the study (appendix 4) on the ward prior to surgery and, if they were happy to take part and met the inclusion and exclusion criteria (Table 7.1), they were recruited into the study at their first rehabilitation session following surgery.

Table 7.1 Inclusion and exclusion criteria for cohort study

Inclusion Criteria	**Exclusion Criteria**
Between the age of 30 and 80	Any neurological impairment leading to lack of comprehension regarding the study or ability to give informed consent
Received TKA surgery on one knee only (at the time of the study)	Any other lower limb impairments (apart from the affected knee) which inhibit normal functional movement
	Unable/unwilling to attend rehabilitation sessions
	Any visual impairment which may prevent the effectiveness of visual feedback
	Participation in any other clinical trial or study
	Contralateral TKA surgery within 18 months leading up to participation in the study
	Any possibility that the participant may be pregnant

Consideration was given to randomisation but given the early nature of the intervention, this was rejected and two patient cohorts were recruited instead. Fifteen patients were recruited into a control group and 15 into an intervention group. All control patients were recruited first, immediately followed by intervention patients. All patients completed baseline and outcome gait tests at the start and end of their rehabilitation. Patient details for each group are presented in Table 7.2. There were no significant differences between groups for any parameters (independent t-test, $\alpha = 0.05$).

Table 7.2 Patient details for each group. Days post-op refers to the amount of time between surgery and the first rehabilitation session

	Controls	**Interventions**
Age (yrs; mean ± SD)	68 ± 6.4	70 ± 6.8
M/F	4/11	4/11
L/R	3/12	11/4
Days post-op (mean ± SD)	15 ± 7	21 ± 18
Days between baseline and outcome testing (mean ± SD)	37 ± 4	39 ± 8

At their initial post-surgery physiotherapy session, patients were asked if they were happy to take part in the study, and providing they were comfortable taking part and remained in accordance with the inclusion and exclusion criteria, a consent form was signed (appendix 5).

Before beginning rehabilitation exercises (appendix 6), all patients completed a baseline gait assessment using the SCM protocol and completed an Oxford Knee Score (OKS) questionnaire (appendix 7). The gait assessment consisted of performing a number of 8 metre long, over-ground walking trials at a self-selected speed, with walking aids if necessary. The number of trials varied between patients depending on number of cycles per trial and individual ability of the patient. Following 6 weekly rehabilitation sessions, all patients also completed an outcome gait assessment and another OKS questionnaire.

The first group of patients were in the control group and therefore completed rehabilitation exercises as normal once a week until 6 sessions had been completed. Appendix 7 details the exercise regime which was followed by all patients.

The second group of patients were in the intervention group and therefore completed rehabilitation with visual feedback. SCM was used to provide feedback to patients for the step-up, STS and weight transfer exercises, the details of which are discussed in chapter 6.

Baseline and outcome gait data were captured using Vicon Tracker and marker trajectories were streamed into D-Flow and filtered using a 2nd order Butterworth filter with a cut off frequency of 10 Hz. Kinematics were then calculated using SCM. Data analysis was performed using MATLAB (Mathworks Inc.) and Microsoft Excel. Data from baseline and outcome trials were stacked and normalised to 100% of the gait cycle using a linear regression fit (appendix 2). Comparisons were made between mean kinematics ± 2 SDs for baseline and outcome data for each patient. Further, baseline and outcome group mean values were compared for four specific outcome measures: peak knee extension velocity in swing (PEVS), peak knee flexion during loading response (PFLR), peak knee flexion in swing (PFS) and gait symmetry (GS) based on peak knee flexion angle. PFLR, PFS were chosen because, visually, these parameters exhibited the greatest change between baseline and outcome gait tests for the majority of patients. PEVS was chosen as this has been proven to be a very sensitive outcome measure (Richards et al., 2003) as it combines the speed at which the movement is occurring and also the magnitude of the movement. Symmetry was calculated using PFS and was chosen to determine if basing feedback goals on the ROM of the non-affected side had a positive effect on gait symmetry. By giving patients goals which were derived from what they could achieve on their non-affected side, it was hoped this would encourage them to achieve the same on their affected side, thus increasing the symmetry between affected and non-affected sides. Symmetry was calculated using the symmetry index (SI; Equation 7.1; Robinson et al., 1987) which suggests that gait is perfectly symmetrical if SI = 0.

$$SI = \frac{|X_L - X_R|}{0.5 \times (X_L + X_R)} \times 100\%$$

Equation 7.1

For each patient, a paired-sample t-test ($\alpha = 0.05$) was carried out to determine any significant difference between baseline and outcome for each outcome measure. The percentage of patients in each group who improved, were less variable and who had a significant improvement for each outcome measure was also calculated.

The mean of all patients means from each group were compared for all outcome measures. Groups were compared using an independent t-test ($\alpha = 0.05$) to determine any significant differences. The majority of patients also completed OKS questionnaires at baseline and outcome to assess improvements in daily functional tasks and pain using a validated scoring method (Clement et al., 2013; Xie et al., 2011). A paired t-test ($\alpha = 0.05$) was used to compare scores at baseline and outcome for each patient and independent t-test ($\alpha = 0.05$) was used to compare the improvement in scores between groups.

7.3 Results

Kinematic data were calculated for all patients; however one representative patient from each group is presented for clarity. Figure 7.2 shows mean baseline and outcome data ± 2SDs for a representative control patient's affected side.

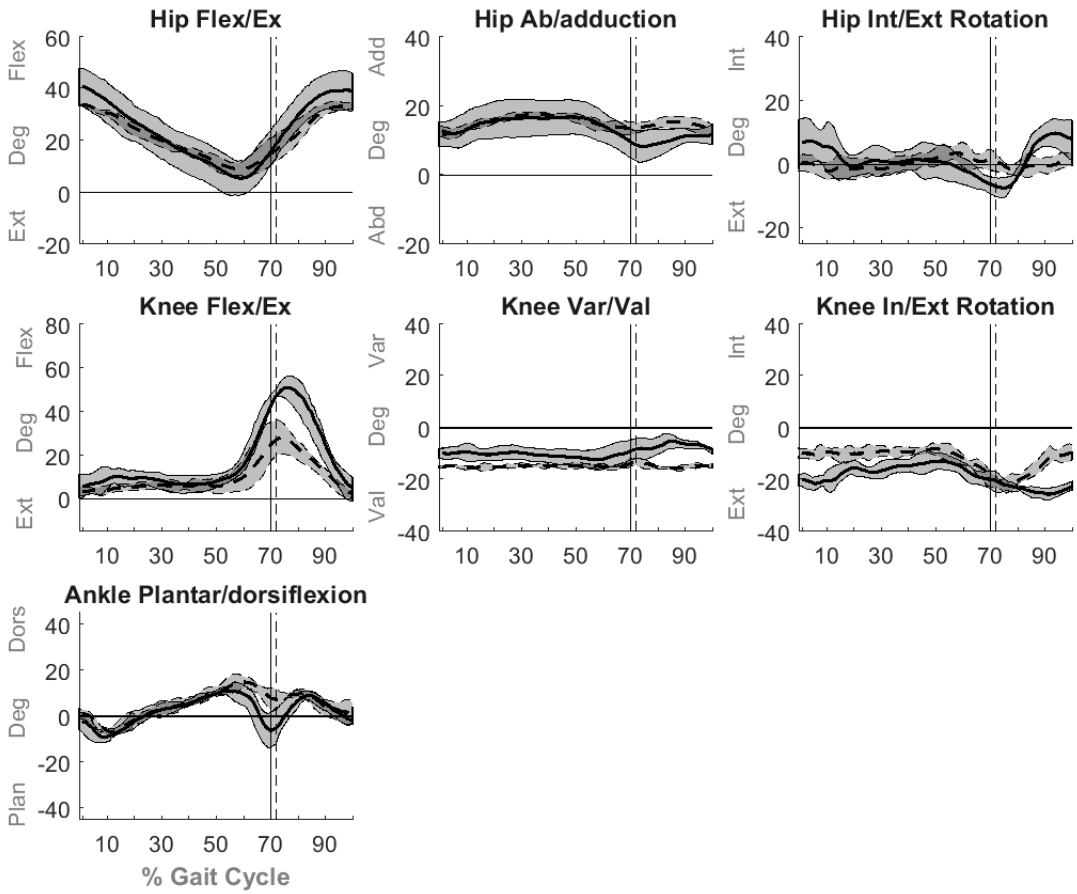


Figure 7.2 Mean kinematic output ± 2SDs for the affected side of one control patient. Mean toe off is represented by a vertical line. Baseline (dashed) outcome (solid)

For hip flexion there was little difference between baseline and outcome; similarly, for hip adduction during stance, baseline and outcome output showed little improvement. However, during swing there was more hip abduction when comparing outcome to baseline and the ROM represented a more normal movement pattern (Baker, 2013). For hip internal/external rotation there was a greater ROM for outcome data compared to baseline, particularly in swing. For knee flexion there was a slight increase during LR and an obvious increase during swing with approximately 20° difference between baseline and outcome. For knee abduction, there was an increase throughout the gait cycle which was most prominent during swing, with an increase of approximately 10°. For knee internal/external rotation the

ROM increased from baseline to outcome. The knee externally rotated for both sets of data; however for outcome data it remained externally rotated for longer in swing, again representing a more normal movement pattern than was seen for baseline data (Czamara et al., 2015; Ferrari et al., 2008). For the ankle, both sets of data were similar during stance; however, during swing, baseline data showed limited plantar flexion at toe-off. Further, the data showed that toe-off occurred earlier for outcome data than baseline data.

Variability was higher for all hip angles for outcome data compared to baseline. Further, variability was also higher for knee abduction and ankle plantar flexion at toe-off.

Figure 7.3 shows mean baseline and outcome data ± 2SDs for a representative intervention patient's affected side.

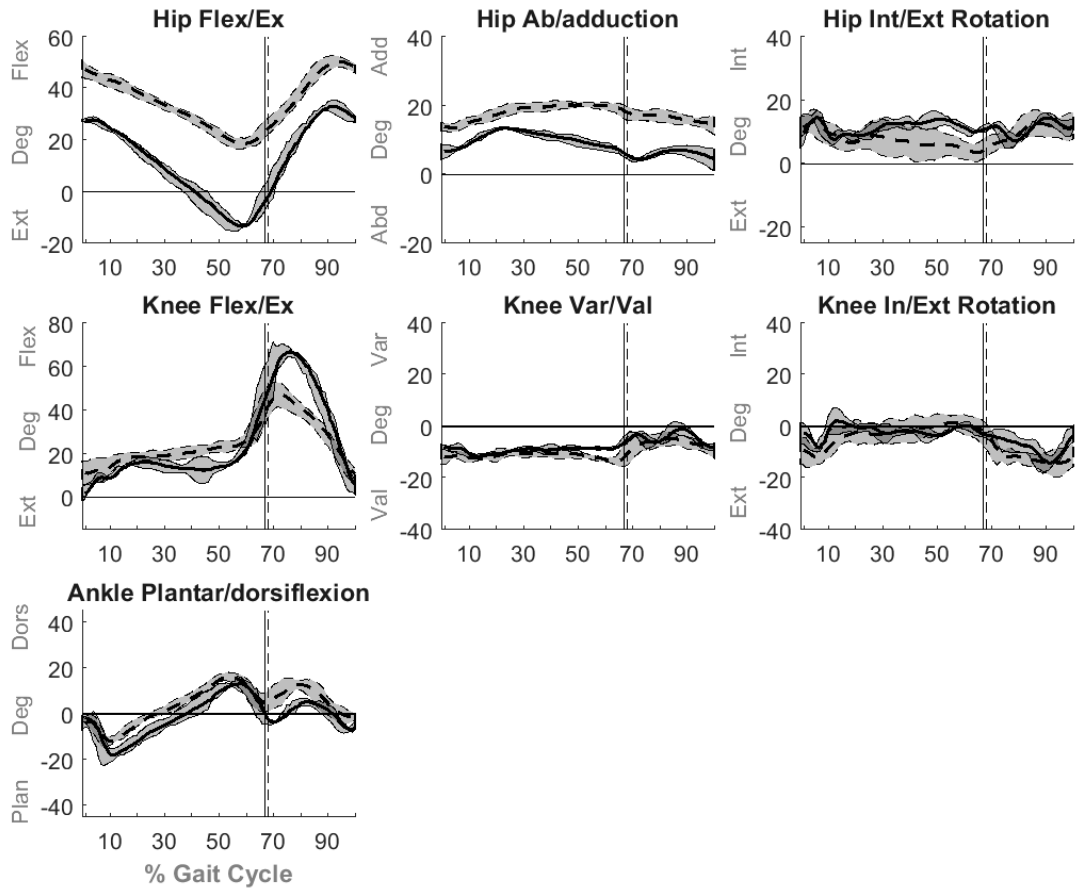


Figure 7.3 Mean kinematic output ± 2SDs for the affected side of one intervention patient. Mean toe off is represented by a vertical line. Baseline (dashed) outcome (solid)

For hip flexion, there was an increase of approximately 20° throughout the gait cycle. Further, the patient achieved hip extension just prior to toe off for outcome data, which is representative of a more normal ROM. There was limited ROM for hip adduction during baseline. However, this increased at outcome with the patient displaying a more normal ROM. For hip internal/external rotation, there was a higher ROM for outcome than baseline, particularly during stance where there was more internal rotation. For knee flexion, baseline data showed limited flexion during LR and swing; however outcome data showed a more normal movement pattern during LR and more flexion in swing. For knee abduction, there were limited differences between baseline and outcome during stance; however there was more abduction during swing for outcome data. There were few differences between knee

internal/external rotation between baseline and outcome data. There was a slight increase in ankle plantar flexion during stance and at toe-off for outcome compared to baseline; however the ROM throughout the rest of the gait cycles remained similar, particularly in stance.

The timing of toe-off was also similar for baseline and outcome; however outcome did occur slightly earlier than baseline. Variability was also similar between baseline and outcome data for all joint angles except hip internal/external rotation, where variability was larger for baseline data.

A number of parameters were also compared at various stages of the gait cycle and control and intervention group data for all patients is presented below.

Table 7.3 shows the mean values for baseline and outcome for each group and each parameter. Means were calculated for each patient and the means were then calculated for each group.

Table 7.3 Mean group values calculated from the mean of each patient at baseline and outcome

	Controls (Mean ± SD)		**Interventions (Mean ± SD)**	
	Baseline	Outcome	Baseline	Outcome
PEVS (°/s)	177.2 ± 73.1	388.7 ± 165.2	233.6 ± 94.2	459.6 ± 132.7
PFLR (°)	14.8 ± 5.4	15.4 ± 5.1	21.1 ± 6.2	20.6 ± 7.1
PFS (°)	42.2 ± 10.7	53.1 ± 10.3	53.1 ± 5.1	63.0 ± 8.3
GS	1.1 ± 1.2	0.7 ± 0.8	0.9 ± 0.6	0.7 ± 0.6

There was an improvement for all outcome measures for both groups except PFLR for the intervention group. For PFLR, the control group saw an increase of 0.6° whereas the intervention group saw a decrease of 0.5° when comparing outcome to baseline. All outcome measures were lower at baseline and outcome for the control group compared to the intervention group, except for GS, which was the same.

Table 7.4 shows the percentage of patients who improved, were less variable and had a significant improvement (paired t-test, $\alpha = 0.05$) between baseline and outcome. All percentages were calculated from mean data from each patient.

Table 7.4 Percentage of patients who improved, were less variable and had a significant ($P< 0.05$) improvement for control and intervention group data. Shaded grey cells indicate which group had a higher percentage of patients

	Controls (%)			**Interventions (%)**		
	Improved	Less Variable	Sig improvement	Improved	Less Variable	Sig improvement
PEVS	93.3	20	73.3	93.3	21.4	92.8
PFLR	66.7	35.7	42.9	40	50	28.6
PFS	86.7	64.3	71.4	93.3	57.1	64.3
GS	60	75	33.3	40	69.2	7.7

For the control group, over 50% of patients improved for all outcome measures. For PFLR and GS more control patients improved than intervention patients and more control patients saw a significant improvement for 3 of 4 outcomes. The outcome measure which resulted in the highest percentage of control patients with a significant improvement was PEVS whereas the lowest percentage was for GS.

PEVS and PFS resulted in the highest percentage of improved intervention patients, each with 93.3% of patients showing an improvement and the lowest was seen for PFLR and GS, each of which saw 40% of patients improving. PEVS showed the highest percentage of intervention patients who had a significant improvement and the lowest percentage was seen for GS, with 7.7% of intervention patients showing a significant improvement.

There was a higher percentage of patients in the intervention group who were less variable for 3 out of 4 outcome measures.

Table 7.5 shows the mean of the mean change per patient for each group.

Table 7.5 Mean change between baseline and outcome for the means of all patients. *significant difference ($\alpha = 0.05$)

	Controls (Mean ± SD; n = 15)	**Interventions (Mean ± SD; n = 15)**	**P Value**
PEVS (°/s)	217.7 ± 150.2	208.3 ± 124.3	0.858
PFLR(°)	0.3 ± 5.2	-0.6 ± 6.7	0.711
PFS (°)	9.8 ± 10.7	12.9 ± 8.9	0.413
GS	0.4 ± 1.3	0.2 ± 0.7	0.421

The control group exhibited a larger change than the intervention group for 3 out of 4 outcome measures; however none of these changes demonstrated a significant difference when compared to changes in the intervention group. The control group saw improvements for all outcome measures; however the intervention group saw a decrease in PFLR.

Table 7.6 shows mean OKS scores and the mean difference between scores for both groups at baseline and outcome. The best score that can be achieved for the OKS is 12 and the worst is 60, therefore a reduction in score indicates a positive result. A paired t-test ($\alpha = 0.05$) was used to determine any significant difference between baseline and outcome for each group, and an independent t-test ($\alpha = 0.05$) was used to determine any significant difference between the change in score of each group between baseline and outcome.

Table 7.6 Mean oxford knee score results for both groups. *significant difference ($\alpha = 0.05$)

	Controls	**Interventions**
N	14	8
Baseline (Mean ± SD)	40.5 ± 6.4	34.8 ± 6.9
Outcome (Mean ± SD)	28.5 ± 7.4	23.9 ± 3.5
P- Value	**<0.001***	**<0.001***
Difference	12 ± 7.2	11 ± 5.5
P-Value	0.75	

More control patients than intervention patients completed the score and controls had higher scores at both baseline and outcome in comparison to interventions. Both groups showed significant improvement in scores after 6 weeks of rehabilitation.

Controls showed a larger decrease than interventions; however, this difference was not significant.

7.4 Discussion

The aim of this part of the study was to determine if rehabilitation augmented with visual feedback had a positive effect on functional outcome for patients who had undergone TKA surgery.

When kinematic data for representative control and intervention patients were compared, there were some similarities. Both representative patients showed an increase in knee flexion during swing, knee abduction during swing, ankle plantar flexion at toe-off and hip abduction during swing. Improvements in hip flexion for the representative intervention patient may be due to reduced pelvic tilt reflecting a more upright stature during gait and therefore a return to a more normal gait pattern. However, this cannot be confirmed due to lack of pelvic kinematics. Future studies should provide pelvic data in order to better determine the mechanism of action or changes in kinematics at the hip.

Toe-off occurred earlier at outcome when compared to baseline for both patients; however, this was still occurring later than might be expected for a healthy age-matched control (Ouellet and Moffet, 2002). Ouellet and Moffet (2002) showed that TKA patients had a larger percentage stance time than healthy controls at 2 months post-op. It was reported that TKA patients achieved toe-off at 68 ± 5.2 % of the cycle compared to 65 ± 2.2 % for healthy controls. These results agree with the findings from the current study, with both control and intervention patients achieving toe-off at approximately 68% which is similar to TKA patients at 2 months post-op. These factors all indicate both representative patients' gait returning to a more

normal movement pattern when compared to baseline (Baker, 2013; Ouellet and Moffet, 2002). This suggests that the rehabilitation had a positive effect on both patient groups in terms of improving functional outcome. In order to determine if one group achieved a better outcome than the other, further investigation into specific gait parameters and differences between baseline and outcome was carried out.

Control patients exhibited lower values for all outcome measures except GS at baseline and outcome compared to intervention patients (Table 7.3). This may have been due to the length of time between surgery and the initial rehabilitation session. Control patients were an average of 15 days post-op at their initial session and baseline assessment whereas intervention patients were an average of 21 days post-op. Patients were instructed to perform rehabilitation exercises at home between leaving the hospital and arriving for their first rehabilitation session so the intervention group had, on average, 6 days more recovery and self-rehabilitation than the control group at baseline testing. This may have contributed to the higher values measured at baseline and outcome for the intervention group.

The reviewed literature revealed limited studies investigating the effects of rehabilitation on gait kinematics for TKA patients at a follow-up of 6 weeks; however, a number of studies have investigated TKA patients' gait at longer follow ups. In the current study, for PEVS at outcome, control and intervention patients achieved 388.7°/s and 459.6°/s, respectively. Jevsevar et al., (1993) aimed to analyse knee angular velocity in patients with TKA and age matched healthy controls. Eleven controls and 10 TKA patients who were at least one year post-op were analysed. Controls achieved an average PEVS of 366.6°/s compared to patients who

achieved approximately 320 ± 45°/s. Brinkmann and Perry (1985) aimed to determine the relationship between gait velocity and rate of motion in healthy subjects and arthritic patients before and after TKA. Seventy-two controls that were not age-matched and were younger than the patient group were tested along with 20 OA patients tested approximately 8 months post-op. The control group achieved a maximum extension rate of 344°/s and the OA group achieved 227°/s.

Richards et al., (2003) aimed to investigate a number of kinematic parameters in stroke patients and healthy volunteers. The stroke patients cannot be compared to patients in this study; however, healthy volunteer data can be used to estimate normal values of knee extension velocity. Ten healthy controls, aged between 65 and 74 were tested and kinematics were measured using an optical based tracking system. Angular velocity was calculated from trials performed at a self-selected pace and the average maximum extension velocity was 386.3°/s.

Outcome results from the control group in the current study are similar to values reported in the previous studies, as they achieved 388.7 ± 63.7°/s. However, results from the intervention group at outcome are higher than values reported in all studies mentioned, both for TKA patients and controls (459.6°/s compared to ~300°/s, respectively). One reason for this could be the measurement methods used to capture data. Jevsevar et al., (1993) used active infra-red light-emitting diodes embedded in rigid plastic arrays placed on the feet, shanks, thighs, pelvis, trunk, arms and head (Figure 7.4a). This setup was bulky and would have likely impinged the subject's movement, which could contribute to the difference seen between results from Jevsevar et al., (1993) and the current study. Brinkmann and Perry, (1985) used a bespoke elctrogoniometer (Figure 7.4b) which they noted demonstrated an average

loss of 6.8° for every 60° of flexion, which could have contributed the difference seen between results.

a. **b.**

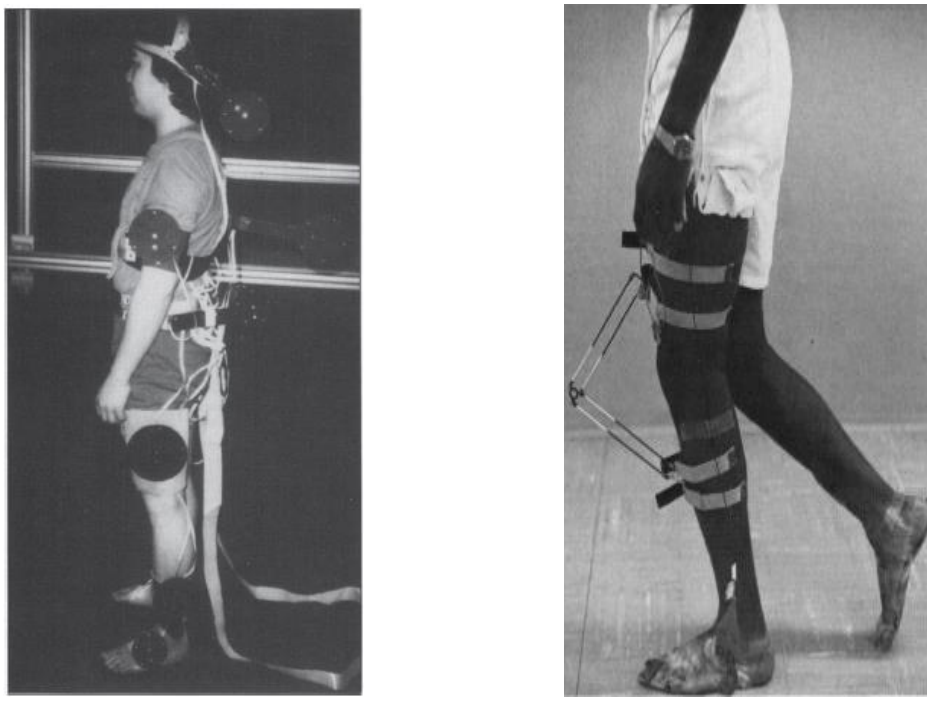


Figure 7.4 Measurement systems used to capture kinematics in **a.** Jevsevar et al., (1993) and **b.** Brinkmann and Perry, (1985)

Due to the limited number of studies investigating knee angular velocity, data from chapter 4, section 4.1.2 was used to calculate peak knee extension velocity for a sample of healthy participants. Knee flexion velocity was calculated for the whole gait cycle using knee kinematics generated by SCM and peak extension velocity in swing was identified. Five participants (mean age = 26.1 ± 3.3) were analysed and the mean peak extension velocity was 475.2 °/s. This value is closer to that seen for intervention patients, and suggests that intervention patients are progressing towards a PEVS value which matches a younger control group.

An improvement in PEVS was seen for 93.3% of patients in each group; however this improvement was only significant for 73.3% of control patients compared to 92.8% of intervention patients.

When the mean of the mean difference per patient was calculated, control patients achieved a greater difference than intervention patients. However, this difference was not statistically significant.

None of the reviewed literature investigated PEVS at various stages of TKA follow-up; therefore results for comparison regarding the improvement in PEVS are limited. However, Brinkmann and Perry (1985) investigated extension rate pre-op and 8 months post-op. They measured a change in peak extension rate of 31°/s, which is lower than results from the current study. Again, this could be due to the use of dated measurement equipment or the amount and/or type of rehabilitation the patients received. There is no mention of rehabilitation so it is possible that patients did not receive any rehabilitation resulting in improvements in PEVS which are lower than those observed in the current study. Pomeroy et al., (2006) stated that a difference in knee angular velocity of greater than 30°/s was thought to be clinically important; therefore suggesting that all patients' PEVS values have improved an acceptable amount.

From these studies it may be suggested that, at outcome, both groups achieved acceptable PEVS values and the change in PEVS between baseline and outcome was greater than that for other TKA patients. At outcome, the control group achieved a PEVS which was similar to other TKA patient groups at various follow-up stages whereas the intervention group achieved a PEVS which was closer to healthy controls. This suggests that provision of rehabilitation has a positive effect on PEVS; however provision of feedback may not result in a higher increase in PEVS in comparison to standard rehabilitation.

For PFLR, the control group saw a small increase between baseline and outcome (0.3°); however, the intervention group saw a small decrease (0.6°). Despite an overall decrease when the mean values were compared, 28.6% of intervention patients still saw a significant improvement in PFLR, compared to 42.9% of control patients. For PFS, both groups improved, with the control group achieving 53.1° at outcome and a change of 9.8° and the intervention group achieving 63° at outcome with a change of 12.9°. PFLR and PFS are two variables which a number of authors have investigated following TKA (Table 7.7)

Table 7.7 Summary of papers investigating knee kinematics following TKA

Authors	Patients/ Controls	Follow-up	PFLR°(mean±SD; patients/controls)	PFS°(mean±SD; patients/controls)
Smith et al., (2006)	34/20	12-18 months	15±5/18±4	54±5/57±4
Brinkmann and Perry, (1985)	20/72	8 months	No data	47±14/62±6
Ouellet and Moffet, (2002)	16/18	2 months	No data	35±8.7/47±5
Benedetti et al., (2003)	9/10	6 months	10±7/16±6	53±8/63±4
Tibesku et al., (2011)	33/none	24 months	No data	58±13
Kramers-de Quervain et al., (1997)	5/none	2-5 years	13±8	52±8
Levinger et al., (2016)	1/3	6 weeks	22/20±9	No data

Follow-up times differ between each paper, and only one paper collected baseline data immediately post-op (Levinger et al., 2016). For all other studies, if baseline data was collected, this was done at the pre-op stage. Therefore, only Levinger's results can be compared directly to this study. However, other results can also give

an indication of whether patients are achieving PFLR and PFS values which might be expected from a TKA patient. In the reviewed literature, PFLR ranged from 10° to 22° for TKA patients and from 16° to 20° for controls. In the current study, control patients achieved PFLR values closer to those seen in the literature at baseline and outcome than intervention patients. Intervention patients generally resulted in higher PFLR values than TKA patients and controls from the literature, However, intervention patients in the current study achieved PFLR values similar to those reported in Levinger et al., (2016) who also received feedback. This may suggest this is an appropriate outcome for patients who received feedback. However there was only one patient in Levinger's study and therefore this is unlikely to be a representative example of PFLR values for patients who have received feedback. Further, when compared to data collected in chapter 4, section 4.1.2 (Table 4.7), intervention patients still resulted in higher PFLR values at outcome with 21.1° compared to 13.6° for healthy participants as measured by SCM. Upon visual inspection of knee flexion curves, it may be suggested that this is because intervention patients are experiencing flexion contractures, which are a common occurrence following TKA surgery (Bong and Di Cesare, 2004; Brander and Stulberg, 2006; Levinger et al., 2016). Figure 7.5 shows mean baseline and outcome data for the affected knee of 3 intervention and 3 control patients. Figure 7.5a demonstrates a mostly uniform increase in flexion throughout the cycle for intervention patients; however, Figure 7.5b shows control patients achieving a small increase in flexion at LR and during swing.

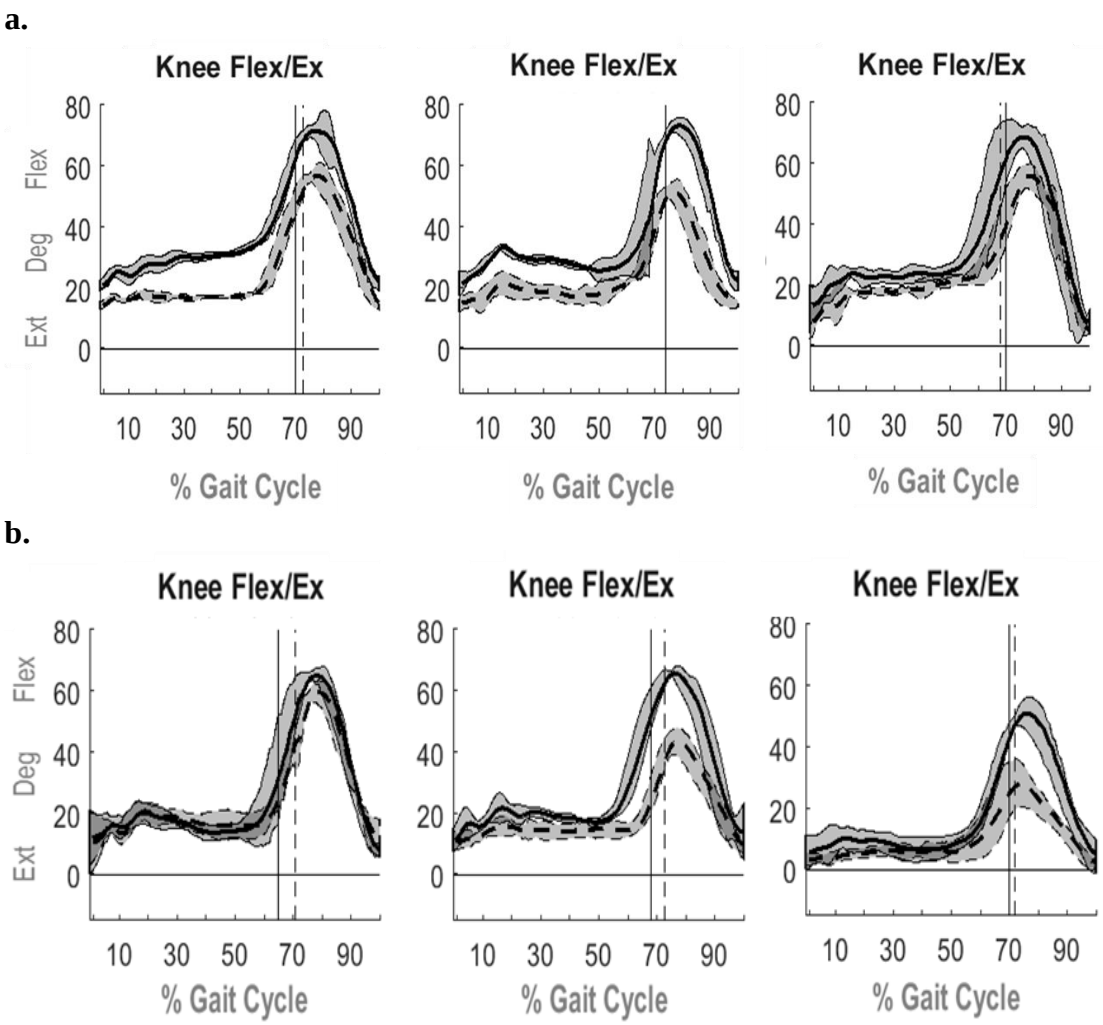


Figure 7.5 Mean ±2SD knee flexion curves for affected knees of **a.** 3 intervention patients and **b.** 3 control patients. Baseline (dashed) outcome (solid)

The uniform increase in flexion for intervention patients is also supported by the PFS values. For PFS, both groups improved, with the control group achieving 53° and the intervention group achieving 63° at outcome. All studies in Table 7.7 assessed peak knee flexion with values for TKA patients ranging from 35° to 58°, compared to controls who achieved 47° to 63°. The results from control patients in the current study are within the upper region of measured values for TKA patients from the literature. However, results from intervention patients are more in accordance with literature values for healthy controls. Further, data from chapter 4, section 4.1.2 (Table 4.7) shows younger healthy participants achieved an average of 55° PFS,

which is lower than the intervention group in the current study. This further supports the hypothesis that the intervention group may be suffering from flexion contractures more than the control group, as TKA patients are unlikely to achieve PFS values higher than younger, healthy controls (Table 7.7). In order to determine if the higher PFS values in the intervention group were due to a uniform increase in flexion, the mean excursion of the knee joint was calculated at baseline and outcome for both groups (Table 7.8). Results demonstrated that the intervention group achieved greater excursion at outcome and a greater increase in excursion between baseline and outcome than the control group, although the difference was not significant (independent t-test, $\alpha = 0.05$, $P = 0.07$).

Table 7.8 Knee flexion excursion angles at baseline and outcome for control and intervention groups. *significant difference ($\alpha = 0.05$)

Knee Flexion Excursion(°)	**Control (mean ± SD)**	**Intervention (mean ± SD)**
Baseline	35 ± 10.1	38.9 ± 9
Outcome	47.5 ± 11.1	55.7 ± 9.4
Difference	12.6	16.7
P-Value	0.07	

This suggests that although some patients may be experiencing flexion contractures, this is unlikely to be the only contributing factor to a larger increase in PFS for the intervention group.

From these results it may be suggested that visual feedback does not have a positive effect on PFLR, but it may have a positive impact on PFS. In future studies, active and passive knee flexion and extension ROM should also be measured to determine if flexion contractures are contributing to the high values seen for PFLR and PFS. However, overall, the mean change was higher for PFS for the intervention group and therefore it may be that the study was not powered enough to detect a statistically significant change.

For gait symmetry (GS), both groups showed an improvement; however, despite both groups achieving the same symmetry index (SI) score at outcome, the control group showed a greater improvement and had a higher percentage of patients who exhibited a significant improvement. Both groups achieved a low SI at outcome, suggesting PFS was acceptably symmetrical at this stage; however, comparison of these results to other studies is limited due to the majority of authors assessing SPT parameters when investigating symmetry (Liikavainio et al., 2007; Patterson et al., 2010). Further, due to the small capture volume used in the current study, PFS values for each side were obtained from different trials, most likely with the patient walking in a different direction. Therefore, GS using PFS was calculated for the group of healthy participants described in chapter 4, section 4.1.1, as this would also require analysis of GS using different trials with participants walking in different directions. Kinematic data calculated by SCM were used to calculate the SI, and the average value for 5 participants was -0.1. Further, Patterson et al., (2010) report the SI value for healthy individuals, calculated using SPT parameters, as ranging from 1.68 to 3.6. The difference between the results obtained in this study and Patterson et al., (2010) may be due to the parameter used to calculate the SI. Sadeghi et al., (2000) stated that parameters which have large values but small inter-limb differences may lower the index and reflect symmetry. However, Becker et al., (1995) successfully demonstrated that surgical treatment of ankle fractures improved gait symmetry in terms of plantar pressure, the values of which are higher than PFS for the knee during gait (Rosenbaum et al., 1994). Therefore, the values reported in the current study are likely to be indicative of adequate symmetry, however, provision of visual

feedback does not appear to have a positive effect on gait symmetry when calculated using PFS.

Both groups' OKS scores increased significantly from baseline to outcome. Clement et al., (2013) stated that the mean improvement for OKS scores is 15 points from pre-op to 1 year post-op. It was also suggested that the minimal clinically important difference was between 4 and 5 points for pain relief and function. Control and intervention patients in the current study saw improvements of 12 and 11 points, respectively from baseline to outcome. This is slightly lower than the value proposed by Clement et al., (2013); however, these scores represent the difference between baseline post-surgery and after 6 weeks of rehabilitation, not pre-op and 1 year post-op. Further, anecdotally, many patients reported a decrease in pain post-operatively compared to pre-operatively, which suggests the difference in scores may have been higher if the score was completed at the pre-op stage. This is supported by Murray et al., (2007) who stated that OKS may be more sensitive when used pre and post-operatively. Control patients saw a larger increase in OKS score than intervention patients which suggests that provision of visual feedback does not have a positive effect on pain and function in terms of activities of daily living. However, this difference was not significant and fewer intervention patients completed scores at baseline and outcome, thus giving less weighting to this group's results.

As mentioned previously, few studies have investigated the effects of TKA rehabilitation with visual feedback. A number of studies have seen positive improvements in function and QoL for stroke patients following rehabilitation with visualisation or virtual reality (Holden, 2002; Merians et al., 2002). However, the

results of the current study are not comparable to these studies as patient groups are different and none of the aforementioned studies focused on gait parameters.

A limited number of studies have investigated the effects of visual feedback on TKA rehabilitation. Levinger et al., (2016) carried out a case study with one intervention patient and 3 controls and used Kinect and Wii to deliver real-time feedback regarding weight distribution, knee flexion angles and body segment alignment during rehabilitation. Results suggested that the intervention patient improved more for external knee extension moment than control patients with changes of 0.7% bodyweight*height (BW*Ht) and 0.03%BW*Ht, respectively. However, only one intervention patient was tested and large SD values (mean = -0.2%BW*Ht, SD = 0.4), were reported for the control patients at the post-op stage, so further study is likely to be necessary to prove the effectiveness of the feedback.

Zeni et al., (2013) tested the effectiveness of a symmetry feedback protocol during TKA rehabilitation. Patients who received symmetry feedback scored higher than controls for the majority of outcome measures at 6months post-op; however it was not mentioned whether any differences were significant. Further, controls were only tested at 6 months post-op, so a comparison of the improvement as a result of the intervention was not possible. However, it was again noted that intervention patients exhibited sagittal plane knee moments similar to normal controls, whereas control patients did not. It was therefore concluded that the symmetry feedback may provide additional benefits beyond those of standard rehabilitation.

Both the aforementioned studies investigated sagittal plane knee moments as an outcome measure and noted that feedback patients returned to a more normal pattern in comparison to controls. Measurement of knee moments was not possible in the

current study and therefore results cannot be directly compared. However both the aforementioned studies and the current study reported interventions performing better than controls for a number of outcome measures. Future studies should consider investigating knee moments as visual feedback appears to have a positive effect on this outcome measure.

A number of authors have also investigated the use of visual feedback in gait retraining in healthy individuals with positive results. Barrios et al., (2010) aimed to test the effectiveness of a knee external adduction moment (KEAM) reducing gait re-training programme. Eight healthy subjects were recruited and 8 sessions of gait re-training were performed where subjects received visual feedback on a screen whilst walking on a treadmill. The coronal plane knee angle was shown and subjects were instructed to lower the stance phase knee adduction angle curves so that they fell within a shaded region on the graph (Figure 7.6).

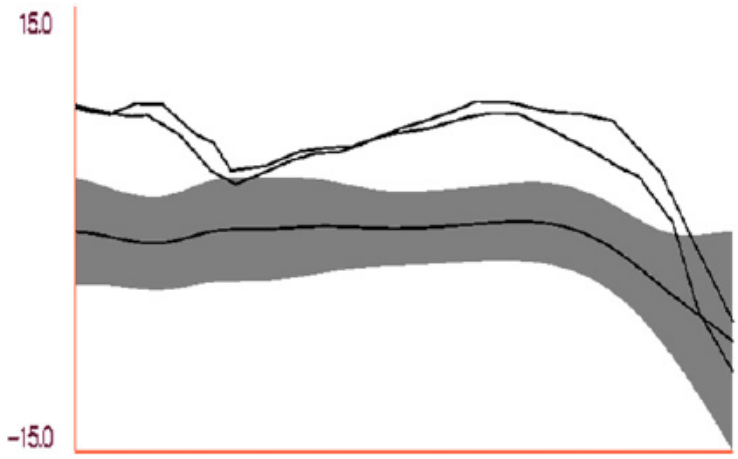


Figure 7.6 Screen image depicting the knee adduction angle data. Shaded area is mean ± 1 SD of normative knee adduction data (Barrios et al., 2010)

Over ground analysis of the subjects' normal gait pattern and their modified gait pattern as a result of the training was performed immediately post-training and at one

month follow-up. At initial analysis and one month follow-up, there was an average of 20% reduction in the KEAM when the modified gait pattern was adopted but the retraining protocol did not alter subjects' natural gait. However, subjects reported that during the training the modified pattern became less effortful to perform and became more natural feeling.

Wheeler et al., (2011) also aimed to present real time feedback of knee adduction moment (KAM) as a method for gait modification to reduce knee joint loads. Sixteen healthy subjects walked on a treadmill with simplified visual feedback of their KAM. Baseline data were collected before training, and outcome was after training but still within the same trial. Results showed that all subjects successfully reduced their KAM by approximately 20% and it was concluded the providing real-time feedback of the KAM was an effective gait retraining method.

Noort et al., (2014) aimed to investigate the effects of various types of real-time visual feedback on the KAM and the hip internal rotation angle (HIR) needed to reduce the KAM in healthy subjects. Seventeen healthy subjects were presented with 4 types of visual feedback regarding either their KAM or HIR (Figure 7.7).

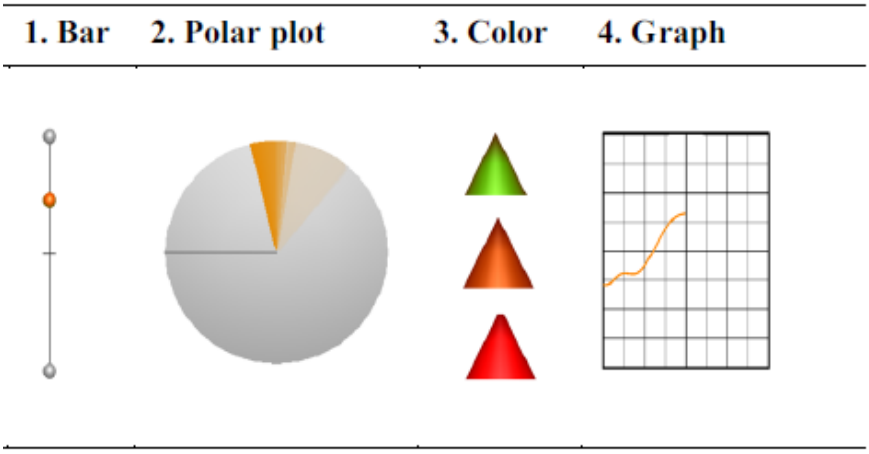


Figure 7.7 Example of the 4 types of feedback patients received regarding their knee adduction moment or hip internal rotation angle (Noort et al., 2014)

Subjects were asked to minimise their KAM or maximise their HIR with respect to their baseline by modifying their gait pattern whilst using one of the 4 types of visual feedback. For all types of feedback, KAM decreased significantly and HIR increased significantly with respect to baseline; however, the type of feedback did not have a significant effect on how much the KAM decreased or HIR increased. Interestingly, KAM did not increase when visual feedback on the HIR was provided. It was hypothesised that an increase in HIR would also lead to a reduction in KAM; however, this was not the case. This suggests that gait alterations may be feedback specific, and if a certain parameter is to be altered, information on that parameter may need to be provided.

From these studies it can be concluded that visual feedback has a positive effect on gait retraining and subjects respond favourably. They also suggest that feedback on the parameters of interest may need to be provided in order to see a positive result. This may be why the intervention patients in the current study didn't perform better than the controls for some outcome measures. The visual feedback focused on improving knee ROM and increasing quadriceps strength and weight acceptance on the affected side, but tested patients on their gait at baseline and outcome. It may be possible that if visual feedback was focused on gait retraining, or active knee ROM was used as an outcome measure, the results may have been more favourable towards the intervention group. If this is the case, one can ask whether visual feedback for TKA patients should be focused on increasing ROM, or should it be more focused on achieving a more normal gait pattern? One hundred percent of patients in the current study stated that their main goal was to return to walking as normally as possible, which suggests that adequate gait is the most important

outcome for the majority of patients. Further study would be required to determine if gait specific feedback and training during rehabilitation had a more positive effect on TKA patients than feedback with standard rehabilitation.

There were a number of limitations with this study. Uptake of visual feedback is completely subjective and dependent on the individual patient. There were some intervention patients who admitted that they didn't use the feedback or didn't pay attention to the screen during exercises. This may have been because the screen which was used to provide feedback was small (Figure 7.8), or that they didn't find the feedback helpful or useful. However, results from chapter 6 suggest that the majority of patients who used the feedback enjoyed it and it helped them complete exercises. Therefore, if a larger screen or a projection onto a wall in front of the patients was used in future studies, this may encourage more patients to use the feedback more.

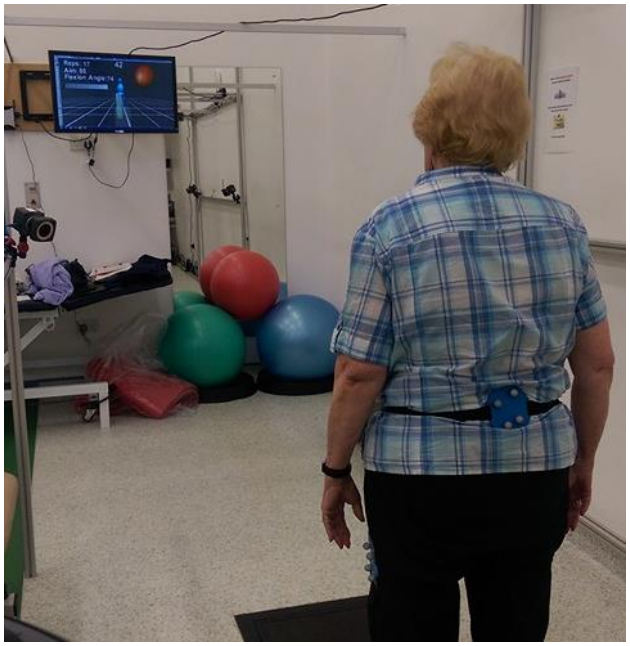


Figure 7.8 Example of a patient performing exercises with visual feedback, the screen used to provide feedback was smaller than was ideal

There were also different lengths of time between surgery and baseline assessment, and baseline and outcome for most patients. Patients were scheduled for their baseline assessment and first rehabilitation class by the MSK centre, as soon as possible after surgery, providing there was space in the class. This resulted in a large range of days between surgery and baseline testing (7 – 37 days), which may have had an effect on baseline and outcome assessment results. This study took place in the hospital and patients were scheduled by the MSK centre. It was therefore not possible to be more strict about time between surgery and baseline assessment or time between baseline and outcome, which also varied depending on whether patients missed a week (30 – 62 days). Future work should aim to structure the study such that time between surgery and baseline, and baseline and outcome testing does not vary, thus reducing the possible impact of this on results.

Further, all aspects of this study were carried out by one researcher, which may have led to bias. However, this could not be avoided as there wasn't adequate time or resource to train a physiotherapist in use of the system and have them run the testing and feedback sessions. One physiotherapist was allocated to run the class, which could contain up to 9 patients at one time, and therefore it was more feasible to have the researcher deal with all aspects of the visual feedback and testing. Throughout all stages of the study, the researcher was careful to avoid bias, and all raw data and results were reviewed by the principal investigator to reduce any chance of researcher bias during analysis.

Finally, the fact that a motion analysis protocol which has not yet been clinically validated was used for gait assessments may be viewed as a limitation to this study. However, data from chapter 4, section 4.1.2 demonstrates that SCM provides a

comparable kinematic output to the current clinical gold-standard (PiG). Further, PiG would not have been appropriate for this study. Due to the high volume of patients who were in the class at any one time, gait assessments were limited to 10 minutes, and setup time for visualisation sessions was limited to 5 minutes. This would not have been possible with PiG, as patients would have needed to change clothing and the application of markers would have been a more complicated and time consuming process. Further, the purpose of the baseline and outcome gait assessments was to show whether there was a difference in gait between these time points, therefore the measurement tool did not necessarily need to provide an extremely accurate representation of gait parameters. It only needed to be able to show whether there was a difference between baseline and outcome testing, which it achieved consistently. Therefore, SCM was considered to be an appropriate model for use in the current study.

Since there were no significant differences for any outcome measures, a sample size calculation was carried out for each outcome measure using Equation 7.2 where α is the confidence interval (0.05), 1-β is the power (0.1; 90% power), *f* is selected from a table depending on the values of α and β, *s* is the standard deviation of the intervention group at outcome (Table 7.3) and δ is the difference between the mean change in the control group and intervention group at outcome (Table 7.5). Results are presented in Table 7.9.

$$n = f(\alpha, \beta) \times \frac{2s^2}{\delta^2}$$

Equation 7.2

Table 7.9 Sample size calculation for each outcome measure which did not show a significant difference in favour of the intervention group. N = patient numbers per group

Outcome Measure	N
PEVS	4185
PFLR	1306
PFS	150
GS	189

Table 7.9 shows that the number of patients required to achieve a significant difference in PEVS, PFLR and GS are likely to be too high to be feasible for a randomised controlled trial (RCT). This is not surprising as for these outcome measures the control group demonstrated a larger difference than the intervention group when the mean of all patient means was calculated. However, a much smaller sample size of 150 patients per group would be required in order to detect a significant difference if PFS was used as an outcome measure, suggesting that this may be an appropriate outcome measure for a large scale RCT.

In conclusion, provision of visual feedback had a positive effect on some outcome measures following 6 weeks of rehabilitation after TKA surgery. Visual feedback did not have a positive effect on PEVS, PFLR or GS. Further study should aim to determine if visual feedback has a positive effect on quadriceps and hamstring strength, and whether feedback for specific outcome measures results in more positive outcomes. Further, knee sagittal plane moments should be considered as outcome measurements in future work as visual feedback appears to have a positive effect on this parameter. The sample size calculation suggests that PFS is also an important outcome measure to include in further study, and as a result of this, and the fact that movement specific feedback may lead to a more positive result, it may be suggested that PFS and active knee ROM should be also used as outcome measures in further studies.

Chapter 8: Clinical Reliability and Acceptability

Chapter 8

8 Clinical Reliability and Acceptability

8.1 Introduction

As has been mentioned previously, the majority of current routine clinical practice relies on observational analysis to assess patient movement (Carse et al., 2013b; Eastlack et al., 1991; Kawamura et al., 2007). However, there has been extensive evidence to suggest that this is unlikely to be the most accurate or effective method for assessing patient progress or outcome (Eastlack et al., 1991; Kawamura et al., 2007; Williams et al., 2009). Limited studies have investigated the reliability of assessing functional movement using observational methods; however a number of studies have investigated observational analysis of gait (Eastlack et al., 1991; Kawamura et al., 2007; Williams et al., 2009).

Williams et al., (2009) aimed to investigate the accuracy of visual observations of gait in patients with traumatic brain injury (TBI) and to determine if experience, type of gait variable, observational plane or body segment had an effect on accuracy of the analysis. Thirty TBI patients and 25 healthy age matched controls were recruited along with 40 assessors who were split into groups: novices (no GA experience), 1st year physiotherapy students with no observational gait analysis (OGA) experience, new graduate physiotherapists with less than 1 year clinical experience, senior physiotherapists experienced in TBI rehabilitation and rehabilitation physiotherapists experienced in TBI rehabilitation. Each assessor observed participants via video footage with direct control over the speed of playback. They were asked to rate 20 gait variables as increased, normal or decreased in comparison to controls. Results demonstrated that assessor inaccuracy for the majority of kinematic variables was

approximately 30-50% and high inaccuracy and high variability was seen for all variables and all assessors. Experienced clinicians were more accurate than inexperienced clinicians for only 8/20 gait variables, with SPT parameters resulting in the highest accuracy. It was suggested that the results indicate the need for objective analysis, particularly in the analysis of gait.

Kawamura et al., (2007) aimed to determine the correlation between OGA and quantitative analysis. Coronal and sagittal plane video recording of 50 cerebral palsy (CP) patients was assessed by 4 physical therapists, each with previous clinical experience in CP plus an additional 2 month training period in normal gait and GA. OGA results were compared between assessors and also to kinematics captured by an optical tracking system (Vicon) using Kappa scores. For between assessors, there was high or moderate agreement for all parameters; however, between OGA and Vicon there was mostly poor or slight agreement. Interestingly, ankle dorsiflexion at initial contact scored the highest kappa value for between assessor analysis (K = 0.88) but scored the lowest kappa value when OGA and Vicon were compared (K = 0.01); therefore suggesting that although all assessors were assigning similar scores for this parameter, they were all assigning scores which deviated from the actual measurement. Further, the high observer agreement was explained by the fact that all observers had taken the same course prior to testing and also had similar professional experiences. It was concluded that OGA cannot be considered as a reliable method.

Eastlack et al., (1991) aimed to investigate the inter assessor reliability of physical therapists' observation based on a videotape of the gait of 3 patients suffering from rheumatoid arthritis (RA) of the knee. Fifty-three physical therapists evaluated 10

gait variables at 4 gait cycle events. Assessors were divided into 2 groups depending on experience; less than or equal to 3 years' experience and greater than 3 years' experience. Agreement coefficients were in the low to moderate range for all variables and it was suggested that this may be because therapists did not seem to be familiar with normative values for the tested gait variables.

From these studies it can be concluded that observational analysis is unlikely to be a reliable and objective method for assessing movement in routine clinical practice. Further, all reviewed studies used video footage which can be paused, rewound and replayed at different speeds, as many times as is necessary. This is not usually a possibility in routine clinical assessment, as clinicians often have to analyse movement in real-time. Therefore, the reliability and accuracy of real-time observational analysis may be further reduced in comparison to video based analysis.

A number of investigators are beginning to use inertial sensors in an attempt to introduce more objective clinical movement analysis without the need for complex motion capture hardware. Hamacher et al., (2014) used inertial sensors to calculate SPT parameters in healthy elderly participants. It was noted that in general, calculation of SPT parameters using sensors is done using double integration of acceleration data, thereby introducing high levels of error into the result and causing issues with reliability and variability. The aim was to address this issue through a repetitive re-calibration process which took place at each stride and determine the reliability after this process had been applied. Nineteen participants were recruited and gait data was collected twice within the first day and once after 7 days. Results demonstrated excellent or good reliability for intra-day testing but fair or poor

reliability for inter-day testing. It was suggested that sensor position between days may have contributed to the low inter-day reliability.

Bautmans et al., (2011) aimed to investigate the reliability of 3D accelerometers to assess SPT gait parameters in a diverse group of young and elderly people. One hundred and twenty-one participants were recruited and included 40 elderly people at risk of falling, 41 elderly controls and 40 young controls. SPT parameters were obtained from one 3D accelerometer placed on the sacrum and same day inter-assessor reliability was tested. When a single walk from each trial was compared, there was fair to poor reliability between assessors. However, when the average of 2 walks from each trial was compared, reliability was good to excellent. It was therefore suggested that multiple walks should be completed per trial in order to ensure good inter-assessor reliability.

These studies suggest that a small number of investigators have demonstrated adequate reliability of sensors to augment routine clinical analysis. However, it also suggests that the majority of sensors are only capable of calculating SPT gait parameters, not kinematics. SPT parameters can provide limited objective information regarding patient progress or outcome; however, they are not capable of providing the type of feedback required during lower limb rehabilitation, therefore are unlikely to be suitable for augmenting the orthopaedic rehabilitation process.

One study (Calliess et al., 2014) developed and validated a specific sensor-based GA system which was designed to assess specific knee outcome measures including SPT parameters and maximum knee flexion angles. The aim was to evaluate the system in a clinical setting by measuring the functional outcome of TKA patients. Walking speed, cadence, step length and knee flexion during normal gait were evaluated for 6

patients. Outcomes were similar to others reported for TKA patients and therefore it was concluded that the system may be a suitable tool for outcome measurement after TKA. However, although previous work validated the kinematic output against 3D instrumented analysis (Schulze et al., 2012), the reliability of the system is not reported and issues with sensor reliability have been reported previously (Hamacher et al., 2014). Further, all software was custom designed and there was no mention of a user friendly interface, which is essential if a system is to be adopted into routine clinical practice.

The reviewed literature suggests that the clinical community is still lacking an objective, reliable and simple method of measuring functional movement to assess patient progress and outcome. There is widespread agreement that instrumented motion analysis is the gold-standard for measuring movement (Cook et al., 2003; Gage, 1993). However, in the clinical environment it is often reserved for complex cases such as patients with CP or other multi-level movement pathologies. This is mostly due to the technical inaccessibility of motion capture technology and the time taken to conduct a session. Historically, the cost of motion capture equipment was a major contributing factor; however, the cost is decreasing and therefore instrumented motion analysis is becoming more accessible (Carse et al., 2013b).

Previous chapters have detailed development of a bespoke, cluster based model (SCM) which aimed to reduce the technical complexity and time taken to conduct a motion analysis session. SCM exhibited a comparable kinematic output to PiG and also displayed overall acceptable reliability with assessors who all had some experience in motion analysis. Further, SCM was acceptable to patients and was successfully used in the clinic by an experienced assessor. However, this is still not

reflective of a realistic clinical environment, as not all assessors will be experienced in motion analysis. In order to be acceptable for routine clinical use, SCM needs to be reliable with clinicians who are unlikely to be experienced in motion analysis. As SCM is a cluster based model, the accuracy and reliability of the kinematic output relies on correct palpation of key anatomical landmarks which may be subject to error between assessors. A number of authors have investigated the difference in anatomical landmark location as estimated by different assessors and the resulting effect on kinematic output (Della Croce et al., 1999; Fukaya et al., 2013; Stagni et al., 2006).

Della Croce et al., (1999) aimed to determine the precision of anatomical landmark determination by 6 physical therapists, all of whom had experience in motion analysis. Results suggested that there were some differences in position of anatomical landmarks between assessors which affected kinematic output. In general, pelvis landmarks resulted in the largest difference in position between assessors. For the kinematic output, the largest difference between assessors for the hip and knee was reported for internal/external rotation and for the ankle the largest difference was for ab/adduction.

Stagni et al., (2006) investigated different calibration techniques and the effect of thigh and shank anatomical landmark misalignment on knee kinematics during 3 functional tasks (flexion against gravity, chair rising/sitting, step up/down). Moderate (5mm) and extreme (15mm) movement of medial and lateral epicondyles markers resulted in differences of greater than 2.5° for all rotations and all functional tasks. The largest difference was 6.8° and was measured during chair rising when

medial and lateral epicondyle markers were moved 15mm and during stepping up when markers were moved both 5mm and 15mm.

These studies suggest that differences in anatomical landmark location between assessors may have an effect on the kinematic output and reliability of a biomechanical model. This effect may be reduced when assessors experienced in motion analysis are tested; however, the majority of clinicians are not experienced in motion analysis. Therefore, the aim of this part of the study was to test the clinical reliability and acceptability of SCM with physiotherapists and podiatrists who have no experience with instrumented motion analysis, in a clinical environment.

8.2 Methods

This study was conducted at the Musculoskeletal Centre, Biggart Hospital, Prestwick, Scotland. One able bodied female subject (age – 27 years, mass – 75kg) and 4 assessors of varying levels of clinical experience (6 – 20 years) who had no experience using instrumented motion analysis took part in the study. All assessors attended a familiarisation session immediately prior to data collection. The session outlined procedures for placement of the clusters and calibration of a participant and each assessor completed a limited number of practice applications and calibrations on the subject prior to data collection. Table 8.1 details the anatomical landmarks which were palpated in order to calibrate the subject.

Table 8.1 Anatomical landmarks which were palpated by each assessor and their corresponding abbreviations

Anatomical Landmark	Abbreviation
Pelvis	
Right Anterior Superior Iliac Spine	RASIS
Left Anterior Superior Iliac Spine	LASIS
Right Posterior Superior Iliac Spine	RPSIS
Left Posterior Superior Iliac Spine	LPSIS
Thigh	
Medial Femoral Epicondyle	ME
Lateral Femoral Epicondyle	LE
Shank	
Medial Malleolus	MM
Lateral Malleolus	LM
Foot	
Calcaneus	CA
First Metatarsal	FM
Fifth Metatarsal	VM
Apex of the First Toe	Toe

Each assessor then applied the SCM markers and calibrated the subject using the methods outlined in chapter 3. Clusters were removed between each assessor. Upon processing it became apparent that there were differences in the position of anatomical landmarks between assessors that would result in significant differences between kinematic outputs. Therefore, the difference between anatomical landmark positions between assessors for the right leg was calculated. This was achieved by using the first assessor's anatomical landmark positions as a benchmark and reconstructing all other assessors' anatomical landmarks in the same reference frame as the first assessor. This was done individually for each segment using the methods for transforming points from one reference frame to another, described in chapter 3. This allowed direct comparison of anatomical landmark positions despite the fact that the subject was not standing in the same position for each calibration. To determine any effect on kinematic output, the same gait trial was processed 4 times, with 4 different calibrations. During each calibration, anatomical landmark positions

for the pelvis and right leg were adjusted to reflect the differences between assessors. Correlation of multiple coefficients (CMC) was carried out on flexion/extension, ab/adduction and internal/external rotation of the right knee to determine agreement between assessors and hence reliability of the SCM with assessors inexperienced in motion analysis.

Further, each assessor was also asked to complete a 5 point Linkert scale for four statements regarding the marker set and calibration process (Table 8.2).

Table 8.2 Statement numbers and accompanying statements for clinician feedback

Statement Number	Statement
1	I found the cluster markers easy to apply
2	Applying the cluster markers was a quick process
3	I found the calibration process easy
4	I found the calibration process quick

8.3 Results

Reconstruction of anatomical landmarks from all assessors revealed some observable differences in position (Figure 8.1). The most obvious differences in anatomical landmark position were observed in the medial epicondyle, whereas the medial and lateral malleolus appeared to be more consistent. Figure 8.1 also demonstrates how much of an effect different pelvis landmark positions can have on the location of the HJC when using a regression equation.

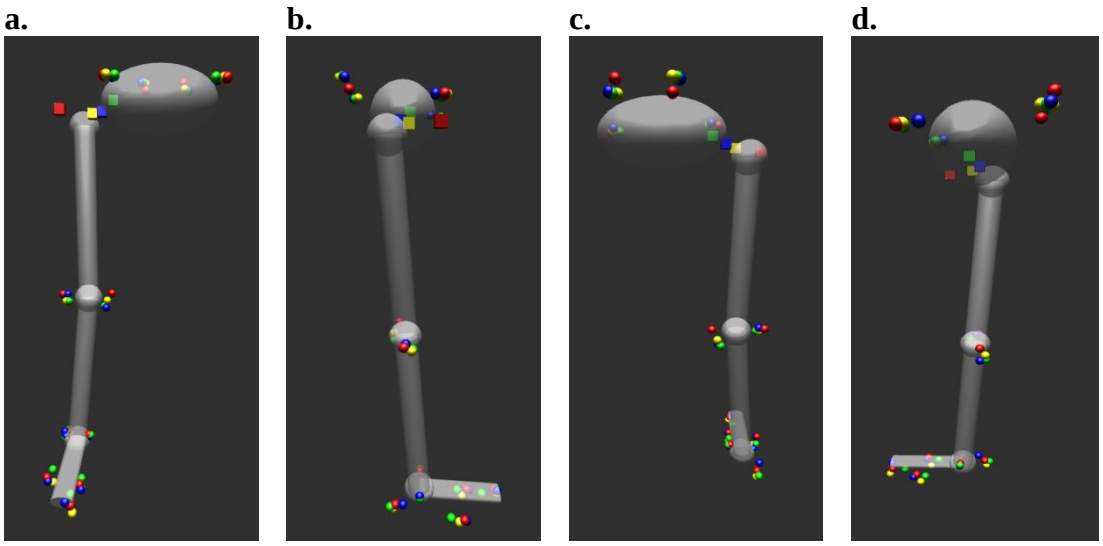


Figure 8.1 Position of anatomical landmarks (spheres) as located by each assessor and the resulting effect on the hip joint centre (cubes). Assessor 1: red, assessor 2: green, assessor 3: blue, assessor 4: yellow. **a.** Coronal (anterior) **b.** Sagittal (lateral) **c.** Coronal (posterior) **d.** Sagittal (medial)

Table 8.3 details the mean ± SD, maximum and minimum Euclidean distance between the HJC and each anatomical landmark between assessors for the right leg.

Table 8.3 Mean(SD), maximum and minimum difference measured between anatomical landmarks and location of the hip joint centre between assessors for the right leg

Landmark	Mean(SD) Difference (mm)	Max Difference (mm)	Min Difference (mm)
R HJC	68.9(34.1)	118.8	26.2
PELVIS			
RASIS	17.1(5.3)	22	6.7
LASIS	21(7.4)	31.1	7.9
RPSIS	23(6.1)	23	6.2
LPSIS	22.5(9)	34.7	5.1
THIGH			
R ME	27.9(11.9)	44.8	14.1
R LE	19.6(7.2)	30.8	8.7
SHANK			
R MM	5.7(2)	8.1	2.5
R LM	7.2(3.7)	13	2.2
FOOT			
R CA	24.2(11.2)	40.6	12.1
R FM	15.5(5.2)	24.1	10.3
R VM	17.8(4)	24.8	12.1
R Toe	18.7(8.5)	31.7	7.2

The smallest mean difference in anatomical landmark position was 5.7mm and was measured at the medial malleolus. The largest mean difference was 27.9mm and was measured at the medial epicondyle. The overall smallest difference was measured at the lateral malleolus and was 2.2mm whereas the overall largest difference was 44.8mm and was measured at the medial epicondyle. For the HJC, the mean difference in position between assessors was 68.9mm, with the largest difference being 118.8mm and the smallest being 26.2mm. Overall, mean differences were lowest for shank landmarks and highest for thigh landmarks.

Figure 8.2 shows the kinematic output of the knee when each assessor's calibration was applied to the same walking trial. Observable differences were limited for knee flexion, however were noticeable for abduction and internal/external rotation, particularly during swing.

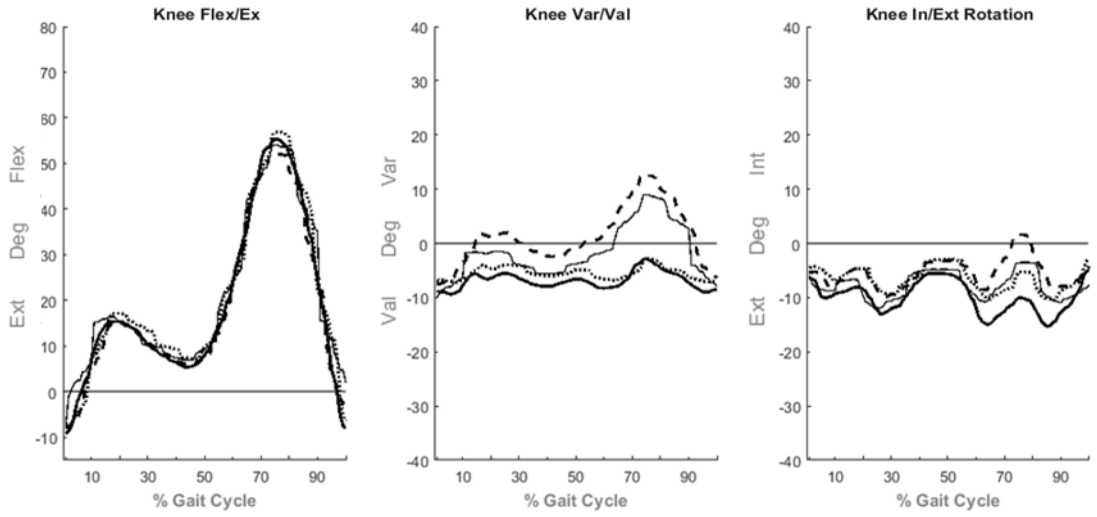


Figure 8.2 Knee kinematics for the right leg using calibrations from 4 assessors. Assessor 1: thick solid, assessor 2: dashed, assessor 3: dot, assessor 4: thin solid

Table 8.4 shows inter-assessor CMC values for knee kinematics of the right leg. CMC values were good for knee flexion/extension and knee ab/adduction and moderate for knee internal/external rotation (Collins et al., 2009).

Table 8.4 Mean (SD) CMC values for knee kinematics of the right leg between 4 assessors

Joint Rotation	CMC Mean (SD)
Knee Flex/Extension	0.98 (0.004)
Knee Ab/Adduction	0.79 (0.13)
Knee Int/Ext Rotation	0.63 (0.19)

Figure 8.3 shows results from the qualitative feedback from four clinicians. The majority of responses were 'strongly agree' or 'agree', although statements 2 (applying the cluster marks was a quick process) and 4 (I found the patient calibration quick) saw 50% of clinicians disagree. Highest agreement was observed for statement 1(I found the cluster markers easy to apply). Statement 3 (I found the patient calibration process easy) also saw high agreement from clinicians. The largest spread of responses was seen for statement 2 (applying the cluster markers was a quick process), as clinicians strongly agreed, agreed and disagreed with this statement.

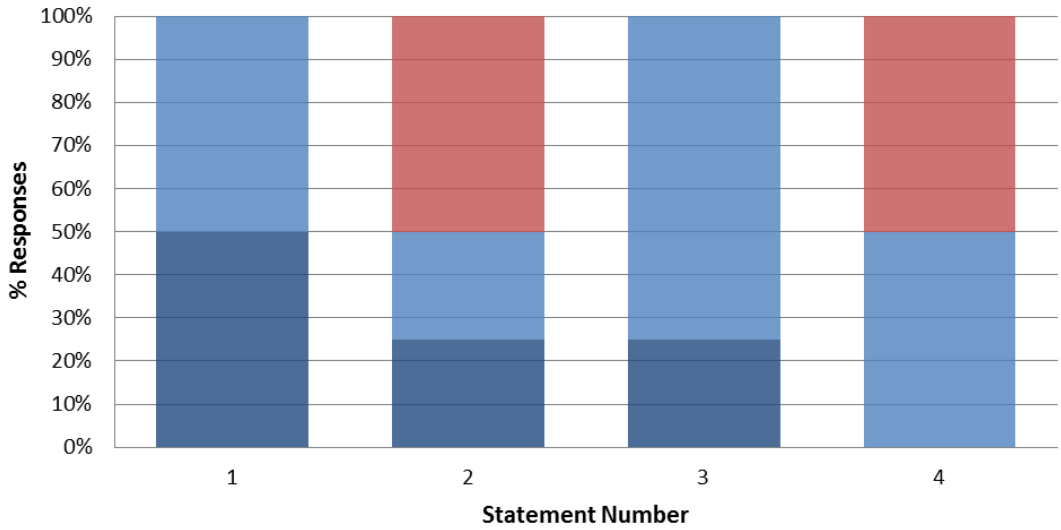


Figure 8.3 Results of clinician feedback. Dark blue: strongly agree, light blue: agree, red: disagree

8.4 Discussion

The aim of this investigation was to determine the reliability and acceptability of the SCM with clinicians who are not experienced in motion analysis. Results of

anatomical landmark identification showed some differences between assessors. The largest Euclidean difference between assessors was 44.8 mm which was recorded for identification of the medial epicondyle. Only one study in the reviewed literature reported values for the difference in anatomical landmark position between assessors (Della Croce et al., 1999). The largest Euclidean distance reported was 24.8 mm and was recorded for the RPSIS which is lower than results from the current study. However, assessors in Della Croce et al., (1999) all had experience of working in a gait laboratory which may have increased the precision of anatomical landmark identification. However, the mean Euclidean distance between assessors for the current study was lower than those reported in Della Croce et al., (1999) for the RASIS, RPSIS, LPSIS, MM, LM and FM which suggests that non experienced clinicians are capable of palpating these landmarks with similar precision to experienced clinicians. In the current study, the medial and lateral malleoli resulted in the smallest difference between assessors (5.7 mm and 7.2mm, respectively), which is lower than Della Croce et al., (1999), who reported mean inter assessor differences of 15.3 mm and 16.8 mm, respectively. This may be due to the clinical expertise of the assessors in each study. In the current study, 3 of the 4 assessors were podiatrists, suggesting that they may be more precise when locating landmarks associated with the ankle. Interestingly, for foot landmarks, 3 of the 4 differences in the current study were greater than those reported in Della Croce et al., (1999), the exception being FM. This may be due to the subject in the current study being shod, making accurate identification of foot landmarks more challenging. The subject in the current study remained shod during testing in order to mimic routine clinical setup as closely as possible. The SCM was used with patients in the clinic for

feedback and assessment of progress during rehabilitation, and the patients were shod during its use. It was therefore important that the reliability testing reflected this. It was not stated in Della Croce et al., (1999) whether the subject was shod or not and therefore it may be possible that the subject was barefoot, thus resulting in more accurate palpation of foot landmarks.

Figure 8.1 highlights the effect of differences in pelvic landmark identification on estimation of the HJC position. The SCM uses the Bell (1989) regression equation which uses the position of the respective ASIS and the distance between the right and left ASIS to estimate HJC location. Therefore, the correct palpation of these landmarks is critical in order to estimate the position of the HJC effectively. The maximum Euclidean distance measured between any two estimated HJC locations was 118.8 mm which is considerably larger than errors reported from other studies using a variety of regression equations and the functional method to estimate HJC position (Stagni et al., 2000). The larger difference in estimated HJC location compared to ASIS positions between assessors was due to the nature of the Bell et al., (1989) regression equation used in this study. Bell et al., (1989) relies on pelvic width and the position of the respective ASIS to estimate HJC position in the pelvic reference frame. Pelvic width is calculated from the position of the RASIS and LASIS landmarks and the largest difference in pelvic width between assessors, as a result of variation in RASIS and LASIS palpation, was 40 mm. This difference had a considerable effect on calculation of the anatomical HJC position within the pelvic reference frame and therefore the global position of the HJC.

It is of concern that a large difference in HJC location could have a significant effect on the resulting kinematics. However, it has been suggested that moving the position of the HJC up to 30 mm in the x, y and z directions has a negligible effect on knee kinematics (Stagni et al., 2000). It was reported that errors in flexion/extension ranged from approximately 0.25° to 1°, errors in ab/adduction ranged from approximately 0.25° to 0.5° and errors in internal/external rotation ranged from approximately 0.25° to 1.25°. The mean difference in HJC location in the current study was 68.9 mm and the maximum difference was 118.8 mm, both of which are larger errors than those investigated by Stagni et al., (2000). Therefore, to determine the effect of HJC location on knee kinematics, the same walking trial as mentioned previously was processed a further 4 times, only changing the position of the pelvic anatomical landmarks and the resultant HJC position to reflect the differences between assessors and determine the effect on kinematic output. Figure 8.4 shows the results of this analysis and confirms that errors in HJC location have less of an effect on knee kinematics than errors in thigh or shank landmarks. The result is less of a change in angle between assessors; however an offset is still present when the difference in HJC location is large. For example, there is 118.8 mm between the HJC position of assessors 1 and 2 (Figure 8.4; thick solid and dashed, respectively) and an offset can be observed in abduction and internal/external rotation between these two outputs.

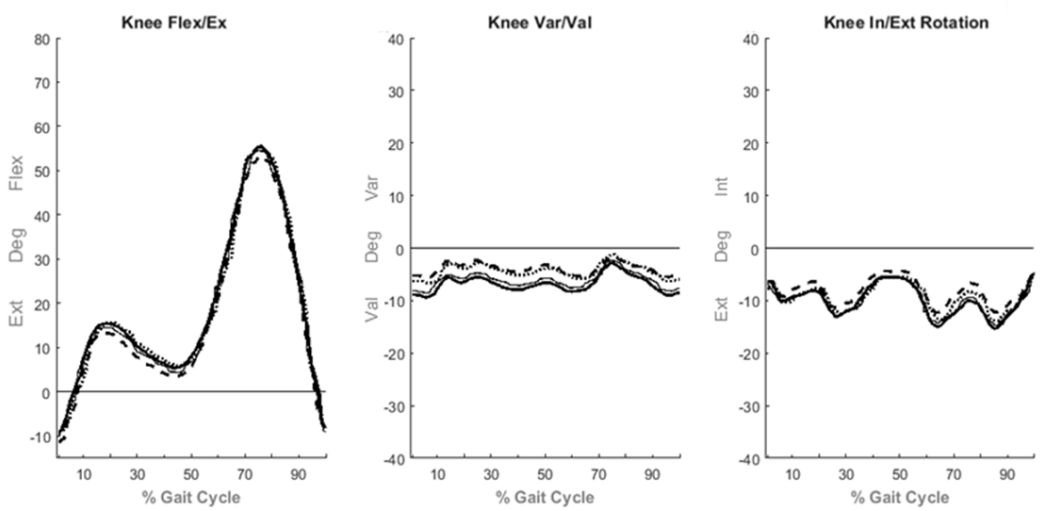


Figure 8.4 Kinematic output for the right knee using calibrations from 4 assessors with changes in pelvic anatomical landmarks only. Assessor 1: thick solid, assessor 2: dashed, assessor 3: dot, assessor 4: thin solid

A number of additional studies have investigated the effects of other lower limb anatomical landmark positions on kinematics (Fukaya et al., 2013; Morton et al., 2007; Piazza and Cavanagh, 2000; Stagni et al., 2006). Piazza and Cavanagh (2000) tested inter assessor variability in palpation of the medial and lateral femoral epicondyles by 5 experienced assessors and the resulting effect on the position of the knee flexion axis. It was determined that the maximum difference between knee flexion axis orientations was 13.5°, which is likely to have a significant effect on kinematic output, particularly abduction as this has been shown to be sensitive to knee flexion axis alignment (Ferrari et al., 2008; McGinley et al., 2009). The difference between LE and ME positions were not stated and therefore the results are not directly comparable to the current study. However, visual inspection of Figure 8.2 indicates that abduction was most affected, particularly in swing, by differences in landmark position. Further, Table 8.3 shows that the ME had the largest mean difference between assessors, which will directly affect the orientation of the knee flexion axis. Therefore, this may be the reason for the large changes in abduction

angle between assessors. The large errors in abduction measured in the current study are also supported by a number of other authors (Della Croce et al., 1999; Fukaya et al., 2013; Morton et al., 2007; Stagni et al., 2006). Della Croce et al., (1999) reported that propagation of errors in knee kinematic output resulting from anatomical landmark misidentification is dependent on flexion. Figure 8.5 shows that as flexion increases, so does the error measured in abduction, which directly compares to results seen in this study where large errors in abduction were measured during swing, when the knee was at maximum flexion.

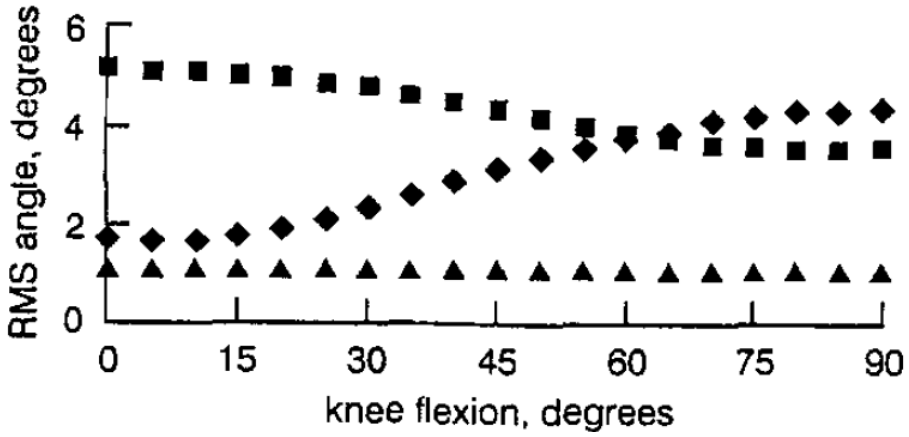


Figure 8.5 Propagation of ab/adduction error in response to increased flexion. RMS – root mean square of the deviation from the mean. Triangles: flexion/extension, diamonds: ab/adduction, squares: internal/external rotation (Della Croce et al., 1999)

Morton et al., (2007) also stated that kinematic results were most sensitive to variability in locating the femoral epicondyles, again with large differences in abduction during swing. Results from these papers and the current study all report high errors in abduction when femoral anatomical landmark position differs between assessors. The combination of these results and results from the current study suggest that misidentification of anatomical landmarks has caused kinematic cross talk between flexion and ab/adduction.

Morton et al., (2007) also stated that kinematic measures with large ROMs were not particularly susceptible to differences caused by variability in locating anatomical landmarks. This also agrees with results from the current study as there is little visible difference in knee flexion kinematics compared to abduction. This suggests that definition of the axes which are used to calculate flexion are less subject to misalignment due to landmark misidentification than those which are used to calculate ab/adduction. In the SCM, the axes used to calculate flexion are the joint coordinate system floating axis and the third axis of the proximal segment (chapter 3, section 3.11), both of which are orientated anteriorly and are formed from the cross product of a longitudinal axis and medio-lateral axis. Therefore, changes in the orientation of the medio-lateral axis will result in differences in the orientation of the floating axis or third axis. As the medio-lateral axis is the flexion axis (the axis about which flexion occurs), and it has been evidenced that this axis is most subject to misalignment due to landmark identification (Ferrari et al., 2008; McGinley et al., 2009; Morton et al., 2007), it may be of concern that this could impact on the floating axis or third axis, thus affecting flexion angle. However, misalignment of the flexion axis would result in a change in orientation of the third axis in the transverse plane, which would not affect calculation of flexion, as this occurs in the sagittal plane. This is therefore likely to be the reason for small changes in flexion in comparison to abduction.

For internal/external rotation, Figure 8.2 shows some differences occurring in swing, similar to abduction, although of lesser magnitude. This is also confirmed by other studies which reported smaller errors in knee internal/external rotation due to landmark misidentification than ab/adduction (Della Croce et al., 1999; Stagni et al.,

2006). Piazza and Cavanagh (2000) suggest differences in internal/external rotation may also be due to kinematic crosstalk. They tested knee flexion and internal/external rotation angles with a correctly aligned flexion axis and with an incorrectly aligned flexion axis using a mechanical linkage. It was determined that an incorrectly aligned flexion axis does result in crosstalk between flexion and internal/external rotation with flexion being misinterpreted as internal rotation when the knee flexion axis was misaligned (Figure 8.6). Therefore, it is likely that the differences in internal/external rotation reported in the current study are also due to misalignment of the knee flexion axis and crosstalk between flexion and internal/external rotation.

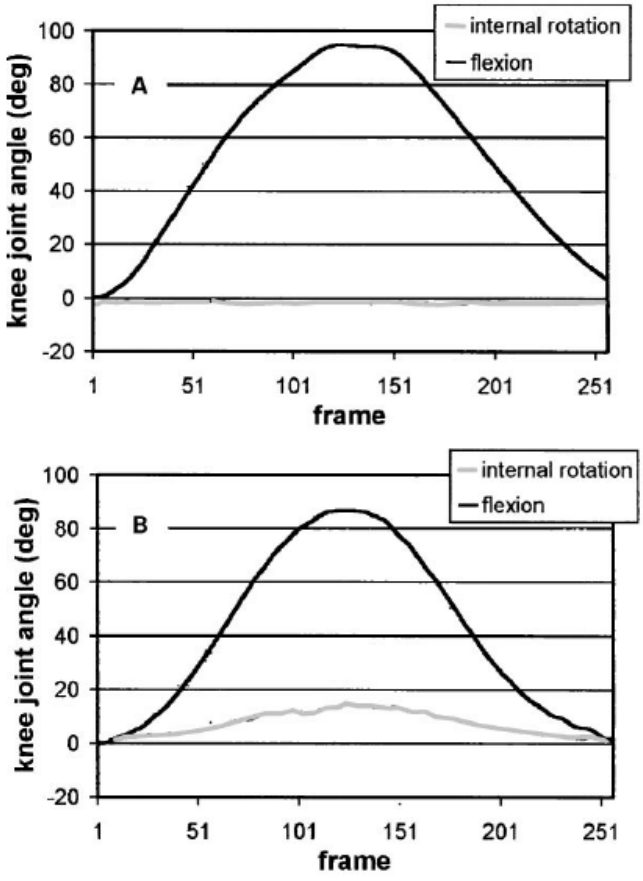


Figure 8.6 Flexion and internal rotation as measured by a mechanical linkage when the knee flexion axis is correctly aligned (top) and incorrectly aligned (bottom) (Piazza and Cavanagh, 2000)

From these studies it may be concluded that the majority of error reported in knee ab/adduction and internal/external rotation in the current study is likely to be due to misalignment of the knee flexion/extension axis due to misidentification of the medial and lateral femoral epicondyle landmarks by some assessors. Although large differences were reported in HJC location and this may be contributing to some of the error reported it is more likely that misalignment of the knee flexion/extension axis is contributing to the large errors.

Despite results from the current study demonstrating differences in knee kinematic output, correlation between outputs was still high or moderate for all rotations (Collins et al., 2009). This is a positive result in comparison to current alternatives for assessing functional movement in routine clinical practice. As mentioned previously, a number of clinicians rely on observational analysis to assess patient progress and outcome (Carse et al., 2013b). However, the reliability of such assessments, even when tested with experienced assessors, is still lower than reliability of SCM with inexperienced assessors (Eastlack et al., 1991; Kawamura et al., 2007). Kawamura et al., (2007) assessed the reliability of observational gait analysis (OGA) between assessors who had specific training in observational analysis. Only 1 of 3 knee parameters resulted in moderate correlation using a kappa test, with the other two resulting in poor correlation (Table 8.5). No parameters were concerned with knee ab/adduction or internal/external rotation and therefore there are no results for comparison.

Table 8.5 Kappa values for knee kinematic parameters observed between assessors Kawamura et al., (2007)

Parameter	Kappa Value
Knee flexion at initial contact	0.54
Knee extension at terminal stance	0.44
Knee flexion at initial swing	0.32

Eastlack et al., (1991) investigated inter assessor reliability of OGA with two groups of physical therapists, one of which had less than or equal to 3 years' clinical experience and the other which had more than 3 years' experience. The results are presented in Table 8.6.

Table 8.6 Kappa values for knee kinematic parameters observed between assessors with varying levels of experience (Eastlack et al., 1991)

	Kappa Value	
Parameter	**≤ 3 Years' Experience**	**> 3 Years' Experience**
Knee flexion at initial contact	0.04	0.26
Knee flexion at midstance	0.36	0.21
Knee flexion at heel-off	0.17	0.25
Knee flexion at toe-off	0.31	0.17
Genu valgum (knee adduction)	0.6	0.52

Results from both these studies demonstrate correlation values which are lower than those reported in the current study. Further, all parameters, except one, are for knee movements in the sagittal plane. This is one of the major limitations of observational analysis as it is only possible to observe movement in one plane at a time and it can be very difficult to connect what is happening in different planes. Therefore, it may be suggested that, despite differences in kinematic output between assessors, reliability of motion capture with inexperienced clinicians is still higher than reliability of experienced clinicians with observational analysis.

Use of accelerometers and gyroscopic sensors was mentioned previously as a possible solution to augment analysis of gait in the clinical environment. A limited number of studies have reported reliability results for use of sensors (Bautmans et al.,

2011; Hamacher et al., 2014); however, the parameters investigated were limited to SPT characteristics and therefore reliability measures are not comparable to the current study. One study (Calliess et al., 2014) did use sensors to calculate knee flexion angle although the kinematic output was not tested for reliability and therefore these results can also not be compared.

From the aforementioned studies, it may be concluded that despite differences in anatomical landmark location and kinematic output, motion analysis is still likely to be more reliable for routine clinical assessment of movement than current alternatives. Further, for use with the lower limb visualisation tool (LLVT) the only measure which is required is knee flexion, which displayed excellent reliability when compared between assessors (CMC = 0.98). This would therefore suggest that for use with the LLVT and for assessment of knee flexion angle, SCM is an extremely reliable method and can be recommended for routine clinical use in this manner. However, if use of the SCM was to be extended to assess outcome using rotations other than flexion, it is advised that assessors receive some training in correct palpation of landmarks. Particular attention should be paid to palpation of the medial and lateral femoral epicondyles and the RASIS and LASIS as misidentification of these landmarks appears to result in the greatest errors in ab/adduction and internal/external rotation kinematics.

Qualitative feedback from clinicians suggests that the most acceptable aspects of the SCM are the ease of use, both in application of markers and calibration of a subject, as all clinicians either agreed or strongly agreed with statements 1 (I found the cluster markers easy to apply) and 3 (I found the calibration process easy). However, 50% of clinicians disagreed with statements 2 (Applying the cluster markers was a

quick process) and 4 (I found the calibration process quick) suggesting that they feel the setup time for the SCM may still be too long for routine clinical use. For statement 4, the other 50% of clinicians agreed and for statement 2 25% strongly agreed and 25% agreed. This may be because one clinician had a limited amount of previous experience attaching the clusters and therefore may have found it faster than the other clinicians. The 50/50 split in agreement and disagreement for statements 2 and 4 suggests that the time taken to setup a subject may vary between assessors, and may be seen as acceptable or unacceptable depending on how much time each clinician has to spend with patients. In this particular clinic, physiotherapists are strictly allocated 20 minutes per patient, whereas podiatrists are allocated approximately 30 minutes and can request longer appointments if they feel the patient may need or benefit from it. The assessors in this study were 3 podiatrists and one physiotherapist and therefore it may be possible that some of the podiatrists felt the setup time was more acceptable than the physiotherapist. Overall, results from the qualitative analysis suggest that future developments of the LLVT should aim to reduce the setup time but retain the ease of use which was highly acceptable to clinicians. This may be achieved by introducing functional calibration of joint centres to the SCM, rather than a pointer based method. This would result in fewer landmarks for palpation and would therefore reduce the setup time.

In conclusion, although there were some differences in ab/adduction and internal/external rotation between assessors, CMC values indicated high or moderate agreement for all joint rotations and reliability for use of SCM as a measurement tool for knee kinematics was higher than current clinical alternatives. Further, clinicians found the SCM easy to use and 50% said they found it quick. Therefore, it can be

concluded from this study that the SCM is acceptable and reliable for routine use to deliver visual feedback regarding knee flexion.

Chapter 9: Discussion and Conclusions

Chapter 9

9 Discussion and Conclusions

9.1 Discussion

Many aspects of this work have been discussed in different chapters throughout this thesis, therefore the aim of this chapter is to combine ideas and themes from all sections and discuss them within the context of the overall aims of the study.

The primary aim of this PhD was to develop a tool to augment the rehabilitation experience of patients and clinicians in a realistic clinical environment. This involved development of a motion analysis protocol tailored for clinical use, use of the protocol to implement real-time feedback during TKA rehabilitation and testing of the effectiveness of the feedback on patient functional outcome. The hypotheses were that routine clinical use of motion capture would become more feasible with a tool which had been designed for purpose and that real-time visual feedback would have a positive effect on patient functional outcome.

This work involved development of a bespoke biomechanical model (SCM) which implemented a pointer based calibration method. This method was shown to be effective for locating anatomical landmarks; however there were some complications which are not present when using static or functional calibration methods. During participant calibration, some cluster markers could become occluded due to another person in the capture volume. Further, cluster markers could also be occluded by the assessor's hand, particularly when palpating the posterior pelvic landmarks, as these were often very close to the pelvic cluster.

When the accuracy of anatomical landmark identification at different pointer orientations was investigated, results demonstrated that the orientation of the pointer

may have a significant effect on the kinematic output of the knee, particularly ab/adduction. This was most likely caused by misidentification of the landmarks which define the knee flexion axis, and has been reported elsewhere (Della Croce et al., 1999; Fukaya et al., 2013; Morton et al., 2007; Stagni et al., 2006). It may therefore be questioned whether pointer based calibration is an accurate enough method for participant calibration, as the pointer will often need to be orientated in different directions in order to reach certain anatomical landmarks. However, the maximum error introduced in one direction when pointer orientation was altered was 5.4 mm, which is less than the majority of errors introduced by landmark misidentification between sessions with experienced assessors (Della Croce et al., 1999). Therefore, when the same assessor is used, any variance in anatomical landmark position caused by pointer orientation is likely to have a small effect in comparison to other sources of error such as intra-assessor variability or soft tissue artefact.

During the calibration process, the hip joint centre (HJC) position was estimated from pelvic anatomical landmarks using a well-defined regression equation (Bell et al., 1989). This resulted in high variation between HJC locations when reliability was tested with clinical assessors who were inexperienced in motion analysis. Results suggested the variation in HJC estimated location did not have a substantial impact on knee kinematics and therefore was unlikely to affect use of the tool for TKA rehabilitation purposes; however, if use of the tool was to be expanded to kinematics at the hip then it is unlikely that this amount of variation would be acceptable. In contrast, functional calibration of HJC position has been shown, in some cases, to be more accurate than regression methods (Leardini et al., 1999) and

may reduce the variation in HJC position between assessors as determination is not dependent on palpation of anatomical landmarks. The current version of the SCM did not employ the functional method as, depending on the specific method used, a lengthy and complicated ROM task may be required (Bell et al., 1990) and this was not deemed appropriate for orthopaedic rehabilitation patients. Piazza et al., (2004) investigated the effects of reduced ROM on estimation of HJC location using the functional method. A number of functional tasks were compared, with an extensive varied hip motion task as the gold-standard. During over ground walking, subjects in Piazza et al., (2004) achieved an average of 46.3° flexion/extension and 14.9° hip ab/adduction, resulting in an average error in HJC location of 50 mm. This is considerably lower than the variation measured in the current study (mean = 68.9 mm, maximum = 118.8 mm) where patients were achieving only a slightly smaller ROM (31.6° flexion/extension and 11.6° ab/adduction of the affected leg at baseline). Further, Piazza et al., (2001) reported mean errors in HJC location of 4.4 mm when hip flexion was limited to 30°. Therefore, while the pointer based method was acceptable and was likely to introduce only small errors into kinematic data, implementation of functional calibration methods into future versions of the SCM using tasks such as over ground walking, may result in lower variation of HJC estimation with clinical assessors. Further, users should be provided with an option to perform static marker based, functional or pointer based calibrations, depending on the capabilities of the patient.

When the kinematic output of SCM was compared to the current clinical gold standard (PiG) there were some small but significant differences. However, the majority of these differences were to be expected. Kinematic crosstalk between knee

flexion/extension and ab/adduction was evident in PiG data, and has been reported previously (Ferrari et al., 2008; McGinley et al., 2009). Differences in the shape of the curve and high variability in PiG data was most evident for knee and hip internal/external rotation outputs which has also been reported previously (Ferrari et al., 2008; Holden et al., 1997; Karlsson and Lundberg, 1994). This work, and previous studies comparing biomechanical models (Collins et al., 2009; Ferrari et al., 2008), is evidence that no two models will produce the same kinematic output, as long as they use different marker sets, different methods to calculate joint centres, different methods to calculate anatomical reference frames or different methods to calculate kinematics. Therefore, it can be advised that, providing the kinematic output of a model is consistent between individuals and over repeated measures, it can be used as a measurement tool. No routine clinical kinematic measurement device will currently be able to provide totally accurate calculation of intersegmental kinematics as this would require the insertion of bone pins. Therefore the most appropriate solution is to offer a measurement tool which is more accurate and objective than current alternatives which mainly consist of observational analysis or manual goniometry. The SCM is capable of providing this with a setup which is much more suitable for routine clinical use than other motion analysis protocols.

However, a number of practical issues persisted when using SCM in the clinic. Although Velcro straps were used to secure clusters as rigidly as possible, movement of clusters did sometimes occur, particularly at the pelvis. This then resulted in clusters having to be repositioned and the patients having to re-calibrated which was inconvenient and time consuming. Future versions of the model should implement

multiple straps on the pelvis cluster and consider using double sided tape on the underside of the plates to reduce cluster movement as much as possible.

As stated previously, a measurement device needs to be reliable if it is to be used for repeated measures analysis. The overall inter and intra-assessor reliability with experienced assessors was comparable between SCM and PiG. However, results of the CMC analysis were lower for hip internal/external rotation for SCM than PiG. This may have been caused by high variability in estimation of the HJC location. Results from chapter 8 highlight the effect a small difference in ASIS location can have on the resultant HJC location, which will have an effect on hip kinematics. Therefore, variability in HJC location in the SCM model could be the reason for low CMC values for both inter and intra-assessor reliability for hip internal/external rotation. However, SCM and PiG use the same regression equation (Bell et al., 1989) to estimate HJC location and the same anatomical landmarks to create a pelvic anatomical reference frame. Therefore similar CMC values might be expected for hip kinematics, although this was not the case as PiG resulted in inter-assessor CMC values of 0.98, 0.96 and 0.94 for hip flexion/extension, ab/adduction and internal/external rotation, respectively; whereas SCM resulted in CMC values of 0.99, 0.93 and 0.53 for the same outputs. It may therefore be possible that assessors were more consistent at locating anatomical landmarks when attaching markers than when using a pointer. Pointer orientation may also have an effect, as it has already been demonstrated that different pointer orientations may result in differences in kinematic output. It has been suggested that short intervals between sessions during reliability sessions may affect results as evidence of marker location may still be present on the participant between assessors (McGinley et al., 2009). This could be a

possible reason for the higher CMC values for certain hip kinematic outputs when PiG was compared to SCM as intervals between assessors were artificially short. However, despite the differences in CMC values for hip internal/external rotation, the overall reliability of SCM was still good, and was certainly higher than other clinical kinematic measurement alternatives (Eastlack et al., 1991; Kawamura et al., 2007). Although reliability analysis demonstrated good results for SCM, it can be difficult to interpret this type of analysis in a practical manner. For example, a CMC value of 0.99 doesn't indicate how different one patient's knee flexion calculation will be from one assessor to the next. Therefore, this type of analysis, while useful to a certain extent, doesn't practically demonstrate the effects of inter-assessor variability. This is more clearly addressed in chapter 8, where the effects of inter-assessor variability can be clearly seen in the position of anatomical landmarks and the changes in kinematic output. However, the question still remains: how reliable is reliable enough? The answer to this is really dependent on what the user is measuring and what they want to get from the analysis. If repeated measures are being used to assess the effectiveness of an intervention, then some degree of reliability will be required. However, a model may prove more reliable for some joint rotations than others, therefore a decision needs to be made about which variables are to be measured and how inter-assessor or inter-session variability affects these outputs.

In terms of the analysis carried out in this study, the accuracy and reliability of SCM was deemed acceptable and it was then used to develop visual feedback scenarios in the form of the lower limb visualisation tool (LLVT). These were then employed in patient rehabilitation and the results regarding patient experience were overwhelmingly positive. There was no report in the reviewed literature of TKA

patient experience when using visual feedback in the clinical environment. However, this is an extremely important outcome measure as patients are far less likely to respond to a tool if they don't like it, understand it, or feel comfortable using it. Results from this study suggest that TKA patients appear to be happy working in a virtual environment and find the use of a motion analysis protocol in their routine care acceptable.

A positive effect on functional outcome in response to visual feedback during rehabilitation has been previously observed in stroke survivors (Carse et al., 2013; Merians et al., 2002). It was therefore suggested that this type of feedback may also improve the efficacy of orthopaedic rehabilitation. However, provision of visual feedback may not have been as effective in orthopaedic rehabilitation as has been demonstrated for stroke survivors due to there being less of a need to re-form neural pathways. Evidence suggests visual feedback is effective when learning a new motor task (Levinger et al., 2016; Subramanian et al., 2010; Swinnen, 1996; Todorov et al., 1997), which is the basis of stroke rehabilitation as the original motor pathways have been destroyed. Some motor control and proprioception is lost after TKA (Pap et al., 2000), however not to the same extent as that following a stroke. Therefore, while visual feedback may be a useful and acceptable addition to orthopaedic rehabilitation, it may not be as effective as has been demonstrated in stroke survivors. That said, visual feedback was successfully and effectively delivered in a routine clinical setting using motion capture, which has never been done before, to the authors' knowledge. Therefore, this may be a promising avenue to continue exploring in order to further improve the efficacy of TKA rehabilitation.

Despite proving to be acceptably reliable with experienced assessors, SCM needs to demonstrate adequate reliability with assessors who are not likely to be experienced in motion capture if it is to be adopted into routine clinical practice. When reliability was tested with inexperienced assessors, CMC values were lower than those observed with experienced assessors for knee ab/adduction and internal/external rotation, but were still higher than values reported in the literature for observational methods (Eastlack et al., 1991; Kawamura et al., 2007). However, the suitability of SCM for routine clinical use doesn't just rely on accuracy and reliability of the measurement system. When clinician opinions were evaluated it was determined that they were still reluctant to adopt use of SCM into routine practice as they considered it to be too time consuming. Therefore, one may ask the question: is it possible to deliver a measurement tool which can maintain the accuracy and reliability of the current version of SCM whilst further reducing patient setup time? Visual feedback scenarios which required kinematics only focused on knee flexion and hence accurate measurement of all joints of the lower limbs may not be necessary to delivery effective feedback. Therefore, one solution could be the option to perform single joint analysis, thus negating the need for a full scale lower limb model for each session. For example, placement of a thigh and shank cluster on one leg, with functional HJC estimation, would allow description of knee kinematics for assessment or feedback. Further, previous evidence has suggested that patients are able to accept seeing only segments of an avatar when receiving visual feedback (MacDonald et al., 2009) and therefore limitation of the model and visualisation to one leg would be unlikely to have a negative impact on the effectiveness of feedback. Consequently, future versions of the SCM should either provide an option for which

segments or joints are to be analysed, or improve cluster recognition so that the software is aware of which segments are in the capture volume at any one time and can therefore determine which segments or joints are to be analysed. One other solution could be the use of a different platform to measure movement which requires less setup time. One study in the reviewed literature (Calliess et al., 2014) used sensors to measure knee flexion. Correlation of knee flexion angle with 3D instrumented motion capture was reported as excellent (r = 0.99; Schulze et al., 2012) and the sensors were used to measure knee flexion during gait. However, the reliability of the sensors for measurement of flexion during gait was not reported and previous evidence has suggested sensors may be particularly subject to low reliability due to double integration of acceleration data and placement variation between assessors (Hamacher et al., 2014). If using sensors to measure knee flexion with multiple assessors could provide similar reliability indices to those measured in the current study then this could be a possible alternative to a fully instrumented 3D biomechanical model for provision of feedback regarding knee flexion. However, the cost of motion capture is reducing and its ease of use increasing, hence while sensors may appear more beneficial in the short term, this may not be the case in the future.

Further, the applications of SCM and the LLVT are broader than just visual feedback of knee flexion in orthopaedic rehabilitation. This study has developed a clinical alternative for reliable and accurate measurement of lower limb kinematics for patient assessment and progress. In order to use everything that the SCM and LLVT are capable of delivering, it makes sense to combine visual feedback and patient assessment. In the current study, over ground gait analysis was used to determine

functional outcome and therefore a combined setup for both feedback and assessment was not possible. However, if a different hardware setup was implemented, for example, motion capture cameras around a treadmill, there would be more options to make use of all the aspects of SCM and the LLVT. Patients could perform feedback exercises, receive visual feedback on their gait and clinicians could perform assessments on either ROM or gait quickly and easily. It is likely that, if all this were possible, the benefits of using the SCM and LLVT would far outweigh the time taken to setup a patient and clinicians would be far more likely to adopt the tools into routine clinical practice. Therefore, by continuing to demonstrate the positive effects of motion analysis in routine clinical practice it may be possible to persuade clinicians that this is a powerful and useful tool which can be used to augment patient rehabilitation and deliver accurate and objective assessments.

9.2 Future Work

There are many aspects of this work which would benefit from further investigation. Continued development of the biomechanical model should aim to remove the dependency on Vicon by including built-in cluster recognition algorithms, thereby eliminating the need for Vicon hardware and software and offering the opportunity to employ cheaper camera systems. Further, the model should implement functional calibration as an option for patients who possess the appropriate ROM. Calculation of moments should also be included in future versions of the model to allow this output for users who possess force platforms or an instrumented treadmill.

Further testing of the effectiveness of visual feedback is also required, as this work only carried out a pilot study. Therefore, further development of visual feedback scenarios should be undertaken to allow more exercises to be performed, and certain

parameters to be altered, such as the difficulty of the task or the amount of information provided. Further, visual feedback applications should be developed for gait retraining and the effectiveness of this intervention should be tested. It may be suggested that further testing regarding the effectiveness of visual feedback should be carried out on patients with neurological deficits, such as those who have suffered a stroke, as these patient groups may respond more strongly to the benefits which visual feedback offers.

9.3 Conclusions

- The accuracy and reliability of SCM are acceptable in comparison to the current clinical gold standard
- SCM is a more suitable model for routine clinical use as patient setup time and technical complexity are greatly reduced in comparison to current motion analysis protocols
- Use of a pointer based calibration method may introduce some small errors into kinematic data; however these are likely to be negligible in comparison to other well documented sources of error
- Patients can accept working in a virtual environment during rehabilitation exercises and respond favourably to visual feedback
- Provision of visual feedback has a positive effect on some outcome measures
- Reliability of SCM with inexperienced assessors is lower than that when tested with experienced assessors, but still higher than current clinical alternatives
- Use of SCM in routine clinical use may still be too time consuming

In conclusion, the SCM and LLVT are acceptable for routine clinical use. Further, provision of visual feedback does appear to have a positive effect on TKA rehabilitation patients; however currently, a full scale biomechanical model may be too time-consuming for clinicians to deliver routine feedback on knee flexion only. If provision of feedback and assessment of progress were combined, the SCM and LLVT would be the ideal tools to provide clinicians with an accurate and reliable method of assessing patients. Future work should focus on furthering provision of feedback beyond the 3 scenarios developed in this study and developing real-time feedback for gait applications. Further, real-time analysis of gait using the SCM and a treadmill should be developed to allow routine clinical gait assessments which can be completed during a patient appointment.

This work has developed a bespoke biomechanical model for routine use in the clinical environment. It has also successfully used the model to deliver feedback in a clinical setting to a larger group of patients than any of the studies in the reviewed literature. Findings from this work indicate the positive effects of motion analysis and visual feedback in orthopaedic rehabilitation and can be used to further the routine clinical use of motion analysis.

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Appendix 1

Index of Electronic Appendices

StepUp
Video of step up feedback scenario

sitToStand
Video of sit to stand feedback scenario

weightTransfer
Video of weight transfer feedback scenario

DFlowAvatar
Code which generates the avatar described in chapter 6

DFlowKinematics
Code which calculates kinematics as outlined in chapter 3

DFlowLoadCalData
Code which loads calibration matrices

DFlowNORMsideStep
Code wish calculates patient goals for weight transfer exercise

DFlowNORMstepUp
Code which calculates patient goals for step up exercise

DFlowNORMsts
Code which calculates patient goals for sit to stand exercise

DFlowPercentageReps
Code which calculates the percentage of reps which were good, medium or bad

DFlowPointerTip
Code which places a marker at the tip of the calibration pointer

DFlowSideStep
Code which executes the weight transfer exercise

DFlowStepUp
Code which executes the step up exercise

DFlowStoreCalData
Code which stores calibration matrices when the button is clicked during subject calibration

DFlowSTS
Code which executes the sit to stand exercise

DFlowTrackLabel
Code which labels cluster markers

FUNCTIONS
Functions which are called within the above scripts can be located in the corresponding 'require' files for each script, saved in the functions folder of the electronic appendix.

Appendix 2

Matlab Code Used to Normalise Gait Cycles

```
function analyseClusterData
global clusterVALID;
```

GET HEEL STRIKE

Get direction of progression

```
zerothPelvisX = clusterVALID.currentData(10,29);
ithPelvisX = clusterVALID.currentData(end-10,29);

if zerothPelvisX < ithPelvisX
    dirOfProg = 1;
else
    dirOfProg = 0;
end

% right
PLz = clusterVALID.currentData(:,29);
rHeelz = clusterVALID.currentData(:,20);

rHeel = sqrt((PLz-rHeelz).^2);

for i = 1:length(rHeel)
    if dirOfProg == 1
        if rHeelz(i,1) < PLz(i,1)
            rHeel(i,1) = eps;
        end
    elseif dirOfProg == 0
        if rHeelz(i,1) > PLz(i,1)
            rHeel(i,1) = eps;
        end
    end
end

[~,clusterVALID.rlocs{clusterVALID.trialNumber,1}] =
findpeaks(rHeel,'MinPeakheight',0.3);
clusterVALID.nrRightCycles(clusterVALID.trialNumber,1) =
(size(clusterVALID.rlocs{clusterVALID.trialNumber, 1},1)-1);
clusterVALID.totalNrRightCycles = sum(clusterVALID.nrRightCycles);

for i = 1:clusterVALID.nrRightCycles(clusterVALID.trialNumber,1)
    clusterVALID.rStartRow{clusterVALID.trialNumber,i} =
clusterVALID.rlocs{clusterVALID.trialNumber,1}(i,1);
    clusterVALID.rEndRow{clusterVALID.trialNumber,i} =
clusterVALID.rlocs{clusterVALID.trialNumber,1}((i+1),1);
end
```

NORMALISE CYCLES

```
xq = (0:1:100)';

% right
if clusterVALID.nrRightCycles(clusterVALID.trialNumber,1) ~= 0
    for i = 1:clusterVALID.nrRightCycles(clusterVALID.trialNumber,1)
        for ii = 1:size(clusterVALID.currentData,2)
            lx = (0:100/(clusterVALID.rEndRow{clusterVALID.trialNumber,i}(1,1)-
clusterVALID.rStartRow{clusterVALID.trialNumber,i}(1,1)):100)';
            lv =
(clusterVALID.currentData(clusterVALID.rStartRow{clusterVALID.trialNumber,i}(1,1):clu
sterVALID.rEndRow{clusterVALID.trialNumber,i}(1,1),ii));
            lvq = interp1(lx,lv,xq);
            tempFullGaitBuild(:,ii,1) = lvq; %#ok<AGROW>
        end
        if ~ isfield(clusterVALID,'rOutputNorm')
            clusterVALID.rOutputNorm = tempFullGaitBuild;
        else
            clusterVALID.rOutputNorm(:,:,size(clusterVALID.rOutputNorm,3)+1) =
tempFullGaitBuild;
        end
    end
% get mean and SD
clusterVALID.MEANrOutputNorm = nanmean(clusterVALID.rOutputNorm,3);
clusterVALID.STDrOutputNorm = nanstd(clusterVALID.rOutputNorm,1,3);
clusterVALID.MEANrOutputNormPlus2sd =
clusterVALID.MEANrOutputNorm+(clusterVALID.STDrOutputNorm*2);
clusterVALID.MEANrOutputNormMinus2sd = clusterVALID.MEANrOutputNorm-
(clusterVALID.STDrOutputNorm*2);
end
```

GET TOE OFF

```
if clusterVALID.totalNrRightCycles ~= 0
    for i = 1:clusterVALID.totalNrRightCycles

        rPLx = clusterVALID.rOutputNorm(:,29,i);
        rToex = clusterVALID.rOutputNorm(:,23,i);

        rToe = (sqrt((rPLx-rToex).^2))*-1;

        [~,clusterVALID.rToeLocs{i,1}] = findpeaks(rToe);
        if numel(clusterVALID.rToeLocs{i,1}) > 1
            clusterVALID.rToeLocs{i,1} = ceil(nanmean(clusterVALID.rToeLocs{i,1}));
        end
    end

    tempMeanRtoeOff = cellfun(@mean,clusterVALID.rToeLocs);
    clusterVALID.meanToeOff = ceil(nanmean(tempMeanRtoeOff));
end
```

Appendix 3

Patients' Questionnaire

Please briefly state anything you liked about the visualisation tool (if nothing please leave blank)
Please briefly state anything you didn't like about the visualisation tool (if nothing please leave blank)

Please answer the following questions stating whether you 'strongly agree', 'agree', 'neither agree nor disagree', 'disagree' or 'strongly disagree' by checking the appropriate box.

	Strongly agree	Agree	Neither agree nor disagree	Disagree	Strongly disagree
I found the cluster markers comfortable to wear					
I found the calibration process (palpation of bony landmarks) comfortable					
I felt comfortable seeing a virtual representation of myself on the screen					
I was comfortable aiming for a virtual target					
I understood the biomechanical information which was displayed on the screen					
I found the biomechanical information useful in helping me to complete the exercise					
I found the visualisations helped me understand how I was moving					
I felt that I could discuss the information on the screen with my physiotherapist					
I enjoyed using the visualisation tool as part of my rehabilitation					

I didn't enjoy using the visualisation tool as part of my rehabilitation					

If you have any further comments about your experience using the visualisation tool please write them in the box below.

Appendix 4

Participant Information Sheet

Study Title: Investigation into the effect of augmenting standard rehabilitation for total knee arthroplasty with visual feedback of functional performance

Researcher: Lindsay Millar
Status: PhD Candidate
Department: Biomedical Engineering
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Chief investigator: Philip Rowe.
Status: Professor
Department: Biomedical Engineering
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Invitation

You are being invited to take part in a study which aims to determine the effect of augmenting the rehabilitation process for patients who have undergone total knee replacement (TKR) surgery. This information sheet outlines why the study is taking place, why you have been asked to participate and what you can expect if you do decide to participate. You are under no obligation to participate in this study. However, if you are interested then please take a few moments to read the information below.

Brief Summary

This study is being conducted by the Department of Biomedical Engineering, University of Strathclyde, in conjunction with NHS Ayrshire and Arran.

We are looking for participants who are undergoing TKR surgery and as a result may need to undergo an outpatient rehabilitation program.

The aim of this study is to determine if visual feedback of movement during rehabilitation has a positive effect on the physical outcome of patients who have had TKR surgery. Currently, the majority of TKR rehabilitation is carried out without visual feedback. Further, assessment of progress and outcome is carried out observationally by the treating physiotherapist which may not be themost accurate method. This study will use motion capture (figure 1a) and visualisation technology (figure 1b) to allow accurate measurement of joint movements and feedback during rehabilitation exercises.

a. **b.**

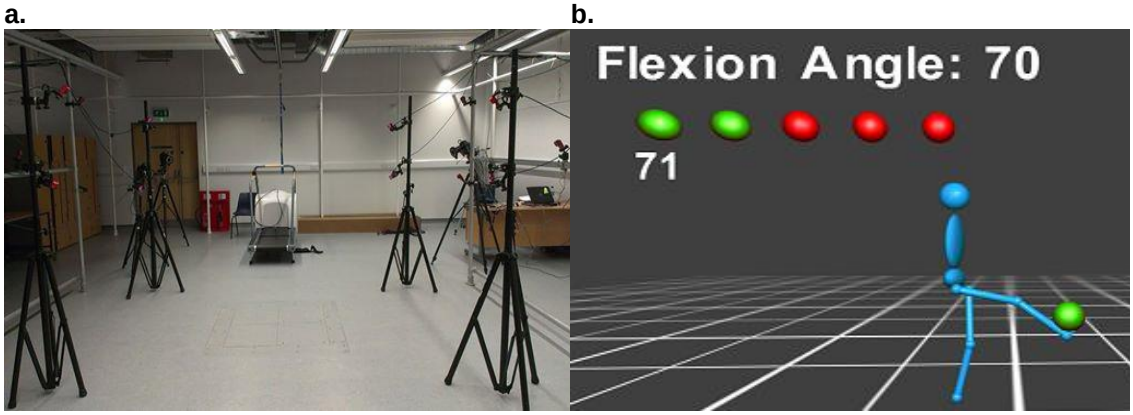


Figure 1a. Motion capture lab setup. Infra-red cameras are mounted on poles and are used to track a participant's movement **b.** Example of visualisation feedback for a leg raise exercise. Movement is visualised with a virtual human or 'avatar'. Virtual targets help to ensure the movement is completed correctly.

After your operation you will attend a 6 week long class which will aim to increase your knee range of motion to above 90 degrees, increase your thigh strength and allow you to walk around independently, without the use of walking aids. Before you begin the class we will assess your movement by asking you to perform a small number of walking trials whilst using the motion capture equipment. Your movement will not be visualised at this time. We will also ask you to fill in a short questionnaire relating to your knee pain and function. After you have completed 6 weeks of rehab, we will assess your movement again and ask you to fill in the same questionnaire regarding your knee pain and function.
This study will require 2 groups of patients. The first group of patients will receive normal rehabilitation with no visual feedback. The second group of participants will use the visual feedback tool for some exercises during rehabilitation. The two groups will be recruited one after the other. The results from the assessments will be compared to see if there are any differences between the feedback and non-feedback groups

Do you have to take part?
You do not have to take part in this study. It is under your own discretion whether you take part in the study. You will have the right to refuse to participate in the study or withdraw from the study at any time without having to provide a reason and without any detrimental effect on your care. You can only take part in this study if you meet the inclusion criteria and exhibit none of the exclusion criteria.

Inclusion Criteria

- You are between the age of 30 and 80
- You have received a TKR on one knee only at the time of the study

Exclusion Criteria

- You have any neurological impairment which means you can't understand why the study is taking place or are not able to give consent to take part in the study
- You have any other lower limb impairments (apart from your replaced knee) which inhibit your movement
- You are unwilling or unable to attend rehabilitation sessions
- You have any visual impairment which may prevent you from benefitting from the visual feedback
- You are currently participating in another trial or study
- You have had your other knee replaced in the last 18 months
- There is any possibility that you could be pregnant

What would taking part involve?
You will not receive any payment or reimbursement for your participation. The study will take place in the Biggart Hospital Musculoskeletal (MSK) centre as part of your routine post-operative care. This research study involves no invasive procedures, and you will not be asked to do any high intensity exercise.

You will be eligible for this study if you meet the inclusion criteria and exhibit none of the exclusion criteria. If you are eligible to take part in the study you will be given a consent form to take home and time to decide if you would like to participate. If you would like to participate, you will undergo a baseline assessment prior to your first rehabilitation session. This will involve attaching a number of 'clusters' of markers to your legs using Velcro straps (figure 2). It will also involve the palpation of a number of bony landmarks to allow the software to estimate the location of your joint centres. The physiotherapist will locate the landmark by pressing firmly on it. This is most accurate when done directly on the skin, so the physiotherapist may have to move your clothing in order to get to the landmark. They will then point at the landmark with a pointer and press a button to allow estimation of the corresponding joint centre; this is called joint centre calibration. Once the button has been pressed, any clothing which was moved can be replaced. Figure 4 shows which anatomical landmarks will require palpation for this study.

a. **b.**

Figure 2a. Plastic plate with 4 reflective markers. You will have a plate attached with Velcro straps to your pelvis, thighs, calves and feet **b.** Example of a female participant wearing the markers.

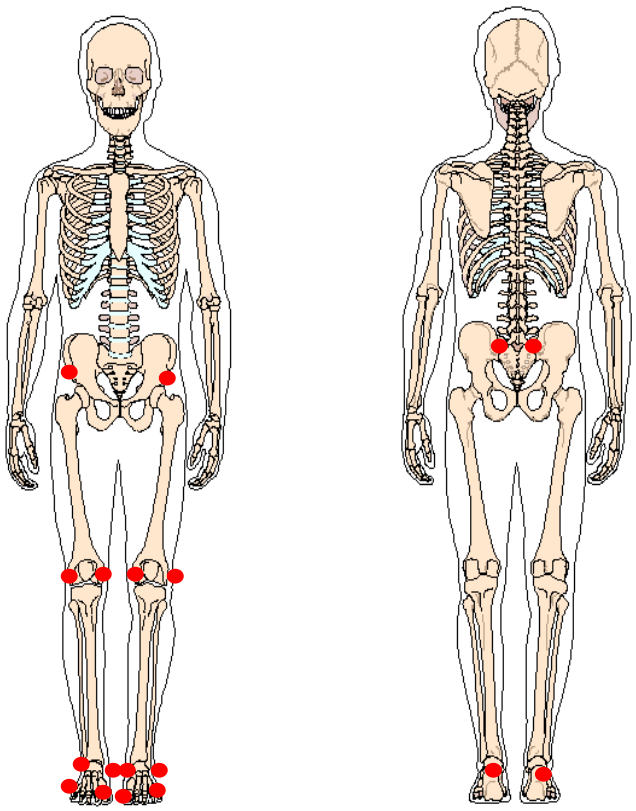


Figure 4. Red dots indicate anatomical landmarks which will be palpated by the physiotherapist.

You will then be asked to perform a number of 8m long walks while wearing the cluster markers at a speed which is comfortable for you while your movement is recorded. The motion capture cameras only record the movement of the markers, you will not be video recorded in this study.

You will then begin your first rehab session and return for your other sessions as scheduled. If you are in the non-feedback group, you will receive normal rehabilitation. If you are in the feedback group you will be asked to wear the same markers as you wore for the baseline assessment and perform some of your exercises using the visualisation tool displayed on a large TV screen, an example of which is shown in figure 1b. The system helps you to see how you are moving and aids the physiotherapist in helping you correct any movement abnormalities or errors.

Once you have received 6 weeks of rehab, all participants will undergo another assessment (outcome assessment), exactly the same as the initial assessment. Once your outcome assessment has been completed no further participation will be requested.

What are the possible benefits of taking part?
Some participants will be receiving an augmented rehabilitation program which allows visualisation feedback of movement during rehabilitation tasks. It is hoped that this added feedback will improve the overall function of the affected knee following 6 weeks of rehabilitation.

What are the potential risks to you in taking part?
This is a very low risk study for participants and there should be minimal risk. However, it is likely that most participants in this study will be unfamiliar with the use of motion capture technology and visualisation tools. The physiotherapists and researcher will do as much as possible to ensure that you have a thorough understanding of what will be asked of you and

where possible, you will be invited to see the motion capture system before taking part if you wish. If you feel uneasy at any point with the use of the motion capture or the visual information displayed on the screen, the visualisations
will be stopped immediately. You will not be asked to do any exercises which would not normally be part of your rehabilitation.

What happens to the information in the project?
All data collected from this study will be treated confidentially and anonymously. Data will be stored on a password locked computer and password protected external hard-drive. Access to the data will be limited to the treating physiotherapist and the study researchers. The results of this study will be submitted for presentation at scientific and clinical conferences and will be submitted for scientific and peer-reviewed publication. You will not be identified in any way.

What happens next?
If you are happy to voluntarily participate in this study then when you come to your first postoperative rehabilitation session we will ask you to complete and sign the consent form on the next page. If you do not wish to participate at this time you just need to tell us and accept our thanks for taking the time to read this information.

This investigation was granted ethical approval by NHS Greater Glasgow and Clyde and the University of Strathclyde Ethics Committee.
If you have any questions/concerns, before, during or after the study then please contact the researcher using the details above. If you wish to discuss the study with an independent person to whom any questions may be directed or further information may be sought from, please contact:

Linda Gilmour
Secretary to the Departmental Ethics Committee
Department of Biomedical Engineering
Wolfson Centre, 106 Rottenrow
Glasgow G4 0NW
Tel: 0141 548 3298
E-mail: linda.gilmour@strath.ac.uk

The University of Strathclyde is registered with the Information Commissioner's Office who implements the Data Protection Act 1998. All personal data on participants will be processed in accordance with the provisions of the Data Protection Act 1998.
Thank you for reading this information – please ask any questions if you are unsure about what is written here.

Appendix 5

Participant Consent Form

Centre Number:

Study Number:

Participant Identification Number for this trial:

CONSENT FORM

Title of Project: Investigation into the effect of augmenting standard rehabilitation for total knee arthroplasty with visual feedback of functional performance

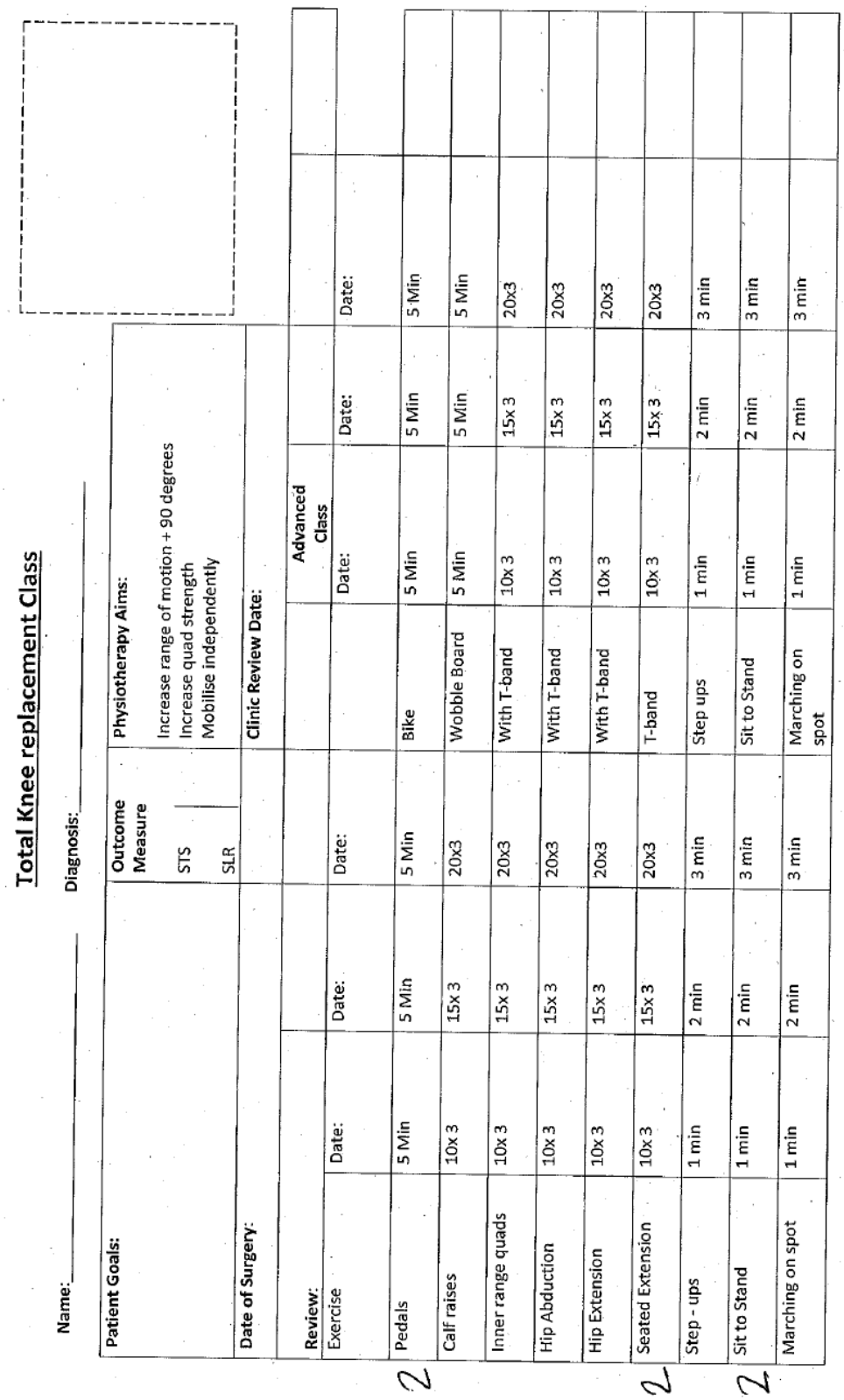
Name of Researcher: Lindsay Millar

	Please Initial Box
1. I confirm that I have read the information sheet dated 14/09/15 (version 1.5) for the above study. I have had the opportunity to consider the information, ask questions and have had these answered satisfactorily.	☐
2. I understand that my participation is voluntary and that I am free to withdraw at any time without giving any reason, without my medical care or legal rights being affected.	☐
3. I understand that the information about me will be used to support other research in the future, and may be shared anonymously with other researchers.	☐
4. I agree to take part in the above study.	☐

_____	_____	_____
Name of Participant	Date	Signature
_____	_____	_____
Name of Person taking consent	Date	Signature

Appendix 6

Patient Rehabilitation Exercises



Total Knee replacement Class

Name:_____ Diagnosis:_____

Patient Goals:			Outcome Measure STS SLR	Physiotherapy Aims: Increase range of motion + 90 degrees Increase quad strength Mobilise independently								
Date of Surgery:				Clinic Review Date:								
Review:						Advanced Class						
Exercise	Date:	Date:	Date:			Date:		Date:				
Pedals	5 Min	5 Min	5 Min	Bike	5 Min		5 Min		5 Min	5 Min		
Calf raises	10x 3	15x 3	20x3	Wobble Board	5 Min		5 Min		5 Min	5 Min		
Inner range quads	10x 3	15x 3	20x3	With T-band	10x 3		15x 3		20x3			
Hip Abduction	10x 3	15x 3	20x3	With T-band	10x 3		15x 3		20x3			
Hip Extension	10x 3	15x 3	20x3	With T-band	10x 3		15x 3		20x3			
Seated Extension	10x 3	15x 3	20x3	T-band	10x 3		15x 3		20x3			
Step - ups	1 min	2 min	3 min	Step ups	1 min		2 min		3 min			
Sit to Stand	1 min	2 min	3 min	Sit to Stand	1 min		2 min		3 min			
Marching on spot	1 min	2 min	3 min	Marching on spot	1 min		2 min		3 min			

2 2 2

Appendix 7

Oxford Knee Score

PROBLEMS WITH YOUR KNEE

During the past 4 weeks.. ✓tick one box for every question

1 *During the past 4 weeks.....*

How would you describe the pain you usually have from your knee?

None	Very mild	Mild	Moderate	Severe
❑	❑	❑	❑	❑

2 *During the past 4 weeks.....*

Have you had any trouble with washing and drying yourself (all over) because of your knee?

No trouble at all	Very little trouble	Moderate trouble	Extreme difficulty	Impossible to do
❑	❑	❑	❑	❑

3 *During the past 4 weeks.....*

Have you had any trouble getting in and out of a car or using public transport because of your knee? (whichever you would tend to use)

No trouble at all	Very little trouble	Moderate trouble	Extreme difficulty	Impossible to do
❑	❑	❑	❑	❑

4 *During the past 4 weeks.....*

For how long have you been able to walk before pain from your knee becomes **severe**? *(with or without a stick)*

No pain/ More than 30 minutes	16 to 30 minutes	5 to 15 minutes	Around the house only	Not at all - pain severe when walking
❑	❑	❑	❑	❑

5 *During the past 4 weeks.....*

After a meal (sat at a table), how painful has it been for you to stand up from a chair because of your knee?

Not at all painful	Slightly painful	Moderately painful	Very painful	Unbearable
❑	❑	❑	❑	❑

6 *During the past 4 weeks.....*

Have you been limping when walking, because of your knee?

Rarely/ never	Sometimes, or just at first	Often, not just at first	Most of the time	All of the time
❑	❑	❑	❑	❑

 /P.T.O

During the past 4 weeks...

✓tick one box for every question

7 *During the past 4 weeks.....*

Could you kneel down and get up again afterwards?

Yes, Easily	With little difficulty	With moderate difficulty	With extreme difficulty	No, Impossible
❑	❑	❑	❑	❑

8 *During the past 4 weeks.....*

Have you been troubled by pain from your knee in bed at night?

No nights	Only 1 or 2 nights	Some nights	Most nights	Every night
❑	❑	❑	❑	❑

9 *During the past 4 weeks.....*

How much has pain from your knee interfered with your usual work *(including housework)*?

Not at all	A little bit	Moderately	Greatly	Totally
❑	❑	❑	❑	❑

10 *During the past 4 weeks.....*

Have you felt that your knee might suddenly 'give way' or let you down?

Rarely/ never	Sometimes, or just at first	Often, not just at first	Most of the time	All of the time
❑	❑	❑	❑	❑

11 *During the past 4 weeks.....*

Could you do the household shopping on your own?

Yes, Easily	With little difficulty	With moderate difficulty	With extreme difficulty	No, Impossible
❑	❑	❑	❑	❑

12 *During the past 4 weeks.....*

Could you walk down one flight of stairs?

Yes, Easily	With little difficulty	With moderate difficulty	With extreme difficulty	No, Impossible
❑	❑	❑	❑	❑