University of Strathclyde Engineering

Evaluation of the errors in kinematic parameters occurring by the use of skin markers in gait analysis, due to soft tissue movements relative to the skeletal system

This thesis is submitted in partial fulfilment of the requirements for the degree of an MSc in Bioengineering

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Declaration of Authenticity

This thesis is the result of the author's original research. It has been composed by the author and has not been previously submitted for examination which has led to the award of a degree.'

Signed:

Date:

Acknowledgements

To everyone who has helped me throughout this process, thank you!

Special thanks go out to my parents and sisters, for their love and support, to my classmates, supervisor and John Maclean for all the help and encouragement, and to Michael for being so patient with me.

1. Abstract

The displacement of soft tissue over the skeletal frame can cause errors when attempting to quantify movements of the underlying bone by using skin markers track human motion. This displacement or "artefact" is responsible for errors in kinematic data acquired by a motion capture system. A method that accurately represents the skeletal image would be beneficial to clinical practice and biomedical research.

In this study, quantification of soft tissue displacement was attempted using two techniques. The first involved a rigid marker cluster that was attached to the shank and thigh of each participant. Skin markers were also adhered to the proximal tibia and lateral sides of the shank and thigh. The change in distances between a cluster and skin marker of each participant during gait was calculated, and this gave a value of soft tissue displacement at each marker. The second part of this experiment involved the use of bi-axial accelerometers, under strapped and unstrapped conditions, that were attached to the same points on the proximal tibia and lateral sides of the shank and thigh. Both results were compared against the body mass index (BMI) of each participant. It was found that the rigid cluster marker system yielded more favourable results than from the use of accelerometers. The results from this comparative study yielded maximal soft tissue displacements of 21, 36 and 68mm in the distal-proximal axis at the tibia, shank and thigh, respectively. It was also found that participant's with higher values of BMI are more prone to soft tissue displacement in regions that contain large amounts of soft tissue.

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CHAPTER 1 - Introduction

1.1. Gait Analysis

Human motion is, in essence, the synchronised effect of the musculoskeletal system. The way a person is able to engage physiological and mechanical means to carry out a specific task reveals a lot about their functional ability. A vast amount of research into the study of human motion has been carried out by a range of scientific branches that include medical, bioengineering, sports science, robotics and space flight programs (Racic et al., 2009). One aspect of the study of human motion is gait analysis. Described in the Oxford dictionary as a “way of walking”, gait involves the execution of movements during the stance and swing phases. Gait analysis is clinically used to determine and address developmental disorders that affect a person’s walk; an uncharacteristic walking pattern can be telling of abnormal or weak aspects of the bone and muscle. In order to correct such abnormalities, a comprehensive representation of the skeletal underlay is vital.

1.2. VICON technology

Several systems exist to assist in human motion analysis, one being the VICON system. VICON technology is a motion capture system that is commonly used in the field of biomechanics. Spherical, reflective skin markers are used to track segment motion, which is captured by a surrounding video graphic system. VICON technology offers a fast, non-invasive, and radiation-free means (Andriacchi et al., 1998) for clinicians to characterise patterns in gait. However, skin and tissue move relative to the underlying bone. The elastic properties of soft tissue give way to the phenomenon known as soft tissue artefact (Leardini et al., 2005). The occurrence of soft tissue artefact means that the desired representation of the skeletal silhouette cannot be correctly distinguished. Thus, the errors arise from the use to skin markers when examining skeletal motion. An additional aspect that can result in errors at the interface between marker and the underlying bone is the misrepresentation of the anatomical landmarks. Anatomical landmarks are known locations on the human body, and these are used to construct an axis in 3-dimensions on the VICON system. The characteristics of soft tissue mean that its viscous and elastic properties vary depending on the location of the tissue.

1.3. Soft tissue artefact

Soft tissue artefact is a significant problem for clinicians and researchers in the field of biomechanics, and is one of the main sources of error in skin motion analysis. Categorizing soft tissue artefact can be done by looking at the regional aspects of the area under examination and investigating the mechanical properties of based on the location.

1.4. Project rationale

To quantify the soft tissue artefact that occurs during gait, a comparative investigation was carried out between two techniques. The first uses the VICON system along with rigid clusters of markers that act as a reference point. The distance between a point on the cluster and a skin marker is analysed and variations to the length can be attributed to soft tissue displacement. The second is the application of skin mounted accelerometers that measure acceleration, performing a double integration to this value gives the displacement at each point. Values acquired through each method will be compared against displacement values obtained by the use of intracortical pins and roentgen photogrammetry (Lafortune and Lake, 1991; Sati et al., 1996). It is hypothesised that values of soft tissue displacement will be higher at areas with denser soft tissue mass than areas with less.

CHAPTER 2. Literature Review

An attempt to quantify erroneous values that arise from the use of skin markers during gait analysis has been addressed, primarily, from assessing the magnitude of soft tissue movement and with techniques to distinguish and locate precise anatomical landmarks.

2.1. Soft Tissue Assessment

A variety of techniques have been implemented to assess soft tissue movement over bone during motion. These include the use of intra-cortical pins, external fixators, percutaneous markers, and Roentgen photogrammetry, ultrasonic technology, accelerometers and a skin displacement sensor. The aforementioned methods indicate that limbs should not be modelled as a rigid body, rather that calculations must be made to quantify and compensate for soft tissue artefact.

2.1.1. Techniques based on intra-cortical pins.

In order to avoid errors due to skin movement, the use of intra-cortical bone pins was introduced. Intra-cortical pins are metal rods that can be surgically imbedded into bony segments as a means of tracking translation and rotation of a limb under observation. In 1948, pioneering researchers (Levens et al., 1948), used intra-cortical pins to investigate the transverse rotation of the segments of the lower limbs in locomotion. Pins were drilled into the upper portion of the tibia, the adductor tubercle of the femur and in the iliac crest of the pelvis. Each pin had an extended rod passing through the tissues and the skin, and on the distal end of the rod there was a spherical reflective marker. Kinematic measurements of 26 subjects were obtained using three synchronised motion picture cameras. Subjects were studied in gait. Results from this study are included in Table 2.1. Captured images revealed that going from the pelvis, to the femur, to the tibia, the angle of rotation increased. On average the range of rotations during gait were 7.7, 15.3 and 19.2°, respectively. It was stated that such a varied range in motion could be due to differences in tissue tension over separate body segments. It could also be due to variations in the individual joints of the various subjects and experimental errors. This study offered information on bone movement during gait however this was an isolated study, the aim of which was to quantify the rotations of the leg along the long axis more accurately than with skin markers.

Table 2.1

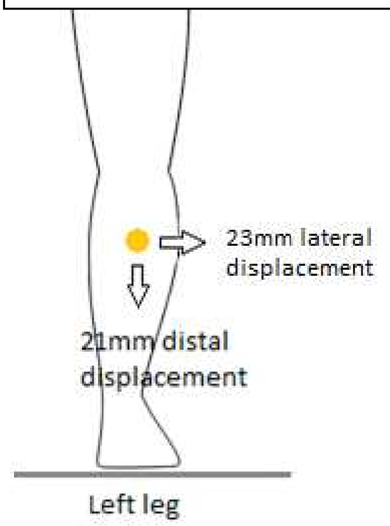
Range of transverse rotations of the pelvis, femur and tibia during gait.

Range of Transverse Rotation			
Subject	*Pelvis relative to the lab axis(°)*	*Femur relative to the pelvis(°)*	*Tibia relative to the pelvis(°)*
8	8	17.6	25.6
11	9.8	9.6	16.4
12	4	10.2	23
13	4.7	15	15
14	10	17.2	19.6
16	7.2	9.8	13.4
21	n/a	24.8	22.8
23	8.4	18	15
24	3	14	17
25	7.4	8.6	17.4
26	9.4	loose	21.4
27	13.3	23.3	24.1
Maximum	13.3	24.8	25.6
Minimum	3	8.6	13.4
Average	7.7	15.3	19.2

Adapted from Levens et al, (1948)

In 1991, the error between skin markers and intra-cortical pins was addressed by Lafortune and Lake (Lafortune and Lake, 1991). Here researchers used pins alongside skin markers to track the bone movements of a subject during gait. X-ray videos along with intra-cortical pins were used to quantify the soft tissue artefact of a subject during gait. Under monitored conditions, subjects were asked to flex and extend the knee cyclically. Findings revealed that a marker placed on the proximal tibia exhibited a 21 mm distal and 23 mm lateral displacement relative to the long axis, which was found to be linearly related to knee flexion. This can be seen in Figure 2.1.

Figure 2.1, adapted from Lafortune and Lake (1991)



In a second experiment in the same paper, soft tissue artefact was analysed at heel strike during a running trial. Data was obtained from a skin marker over the lateral tibial condyle and a cortical pin inserted into the tibia of one volunteer. Findings revealed that relative movement between the marker and the pin reached 10 mm. It was noted that the nature of impact dictated the magnitude between the two markers.

A study that followed in 1994, detailed how Karlsson and Lundberg used external marker devices along with skin markers to track skeletal hip rotations. Each external marking system consisted of a bone screw and an aluminium tripod measurement apparatus with three reflective spherical markers. One was attached to the distal femur and another to the proximal tibia. Three skin markers were also taped to the distal thigh and proximal shank. Two subjects were asked to perform hip internal-external rotation with the knee locked while standing. It was found that there was a large discrepancy between knee joint rotations obtained with bone anchored and skin attached markers. A depiction of this is shown in Figure 2.3. Rotations of 20° were noted from the external marking system and rotations of 50° for skin attached markers. Markers on the shank displayed a smaller displacement result than those on the thigh.

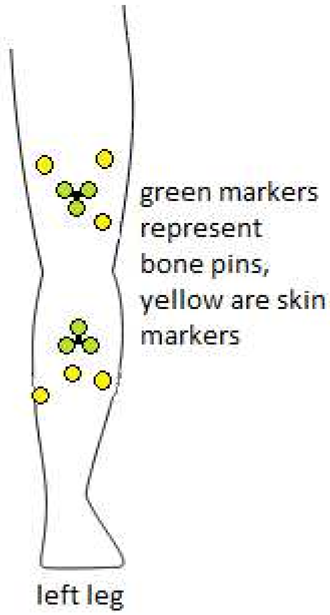


Figure 2.3, adapted from Karlsson and Lundberg (1994)

Reinschmidt et al. (1997) assessed soft tissue contribution both in knee and ankle motion during walking. Intra-cortical assessment Hoffman pins with reflective markers were inserted into the lateral femoral condyle, lateral tibial condyle and posterior-lateral aspect of the calcaneus in three volunteers. Six skin markers were also adhered to the thigh, six on the shank and a further six on the shoes. Segmental error analysis confirmed that most of the error in knee flexion/extension was due to the soft tissue artefact at the thigh. A technique that provided a simple geometric description of the joint coordinate system with the three-dimensional rotational and translational motion between two rigid bodies was used. The study concluded that skin markers should only be used to determine flexion-extension at the tibiofemoral joint. Values of this study can be seen in Tables 2.2 and 2.2.1. The root mean square value of each rotation was calculated. In mathematics, the root mean square (RMS) is a statistical representation of the average value of a varying result. Results showed that skin displacement at the knee during abduction-adduction motion and internal-external rotations was as high as the real joint motion.

Table 2.2

The root mean square differences (RMS diff) and maximal difference (Max. diff) between bone and skin based markers for rotations at the knee.

Rotation at the knew between proximal tibia and distal femur	*Variable (°)*	*Sub.1*	*Sub 2.*	*Sub 3.*
Ad/abduction	*RMS diff.*	2.1	2.4	2.8
	Max. diff	3.1	4	6
Int./ext. rotation	*RMS diff.*	4.2	2.1	5.3
	Max. diff	7.6	7.3	10.3
Flexion/extension	*RMS diff.*	1.5	1.7	3.2
	Max. diff	2.6	4.6	5.8

Adapted from Rienschmidt et al. (1997)

Table 2.1

Root mean square differences and maximal differences between bone and skin based for knee and ankle complex rotations in the three anatomical planes during stance phase of walking for three healthy subjects.

Anatomical planes	*Variable (°)*	*Knee rotations*			*Ankle complex rotations*		
		Sub.1	*Sub.2*	*Sub.3*	*Sub.1*	*Sub. 2*	*Sub. 3*
Frontal	*RMS diff*	2.1	2.4	2.8	4.4	3.6	2.9
	Max diff	3.1	4	6	6.4	5.6	4.2
Transverse	*RMS diff*	4.2	2.1	5.3	4.3	2	3.2
	Max diff	7.6	7.3	10.3	5.7	5	4.5
Sagittal	*RMS diff*	1.5	1.7	3.2	3.1	4.4	2.5
	Max diff	2.6	4.6	5.8	4.9	8.1	4.6

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Adapted from Reinschmidt et al. (1997).

The same authors investigated the effect of soft tissue movement over joint motion during five running trials (Rienschmidt et al., 1997). Cardan angles were taken from both skin and skeletal markers. From the calculation of the Cardan angles, is it possible to locate rigid human body segments in 3-dimensional space (Tupling and Pierrynowski, 1987). Findings showed that upon comparison of skin and bone markers around the knee for extension-flexion, there were few discrepancies. Errors reported were 21% relative to the full range of motion. However, when comparing abduction-adduction motion and internal-external rotations, the discrepancies were much higher. Errors reported reached 64% and 70%, respectively. In this study, it was stated that skin markers can lead to an over estimation of joint motion. In addition to this, it was found that skin movement errors are higher in running than walking.

In 1997, Fuller et al. implemented a leg with two assemblies of six markers that were directly inserted into the tibial tubercle and greater trochanter. An additional twenty skin markers were evenly adhered over the thigh and shank segments. A single volunteer was asked to perform various motor tasks, and results obtained revealed that displacement in skin markers and bone markers differed by up to 20 mm. Fuller et al. (1997) found that different tasks determined the amount of soft tissue movement. Researchers of this study concluded that skin markers should not be used to skeletally track a subject's movement in gait, particularly around the femur.

A paper (Yack et al., 2000) described the use of a device that consisted of intra-cortical pins, which had been fitted with four infra-red light emitting diodes. These devices were surgically fitted into holes drilled into the proximal-lateral aspect of the right femur and tibia. Three skin mounted markers were also placed on the crest of the tibia. The femur was tracked using an infra-red pin device along with two diodes that were attached to its lateral extension. Two volunteers were asked to perform running and walking tasks; these movements were analysed. Results revealed that an erroneous skin displacement over the tibiofemoral segment had the root mean square value of 10mm**.**

A subsequent paper from the same authors (Mouke and Yack, 2004) compared the tibiofemoral motion during gait with the femoral tracking device (FTD), described in the previous study, and bone mounted markers (BMM). In the same paper, authors also compared tibiofemoral angles and displacement during walking. Set up of this experiment can be seen in Figure 2.3.2. Results from this study can be seen in Table 2.3. The absolute differences between the femoral tracking device (FTD) and the bone mounted markers (BMM) were caluculated by the author to be on average 5.3 ± 2.1 □ and maximally 14.2 ±5.1 □. It was concluded that minimal rotational errors can be obtained using an infra-red diode device over bone anchored markers. An acceptable range of error is agreed to be within a few degrees**.**

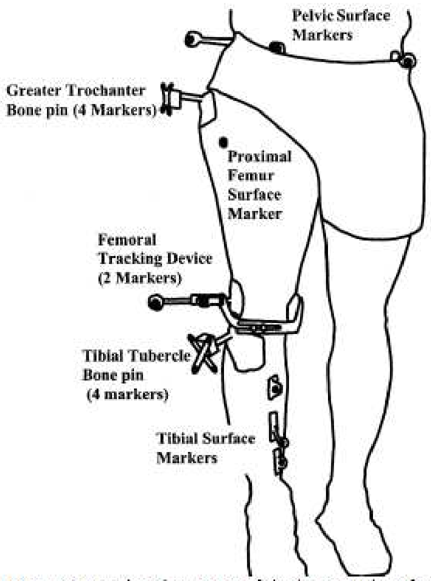


Figure 2.3.2 The placement of the bone and surface mounted markers are illustrated here. Adapted from Houke and Yack (2004)

Table 2.3

The average and maximum tibiofemoral displacement for each subject for the first 85% of stance using the Femoral Tracking Device (FTD) and Bone Mounted Markers (BMM).

	FTD (mm)		*BMM (mm)*		*Absolute differences (FTD-BMM)*	
Subject	*Ave*	*Max*	*Ave.*	*Max*	*Ave.*	*Max*
A	2.2	3.4	11.9	18.8	9.7	16.3
B	3.4	7.3	7.2	13.2	3.9	11.8
D	4.8	7	5.6	14	2.3	7.8
E	6.8	10.8	11.4	25.7	4.7	15.2
G	4	10.1	7.6	24.2	4.8	20.5
I	3.8	6.5	10.1	16.3	6.3	12.1
L	4.8	9.4	9.5	11.8	5	7.4
N	6.4	13.3	14.8	23.5	8.5	17
P	3.7	5.9	8.7	20.2	5.6	15.1
R	5.1	8.8	8.1	11.2	3.1	8.9
S	6.1	8.7	98	19	4	14
T	3.7	9.2	9	28	6	24.7
Ave.	4.6	8.4	9.5	18.8	5.3	14.2
Max	6.8	13.3	14.8	28	9.7	24.7

Adapted from Yack et al. (2004)

Ankle-joint complex motion during stance phase of walking was measured using skin and bone anchored markers (Westblad et al., 2000). Assessment was carried out by measuring the skin and bone anchored markers in three volunteers. Hoffman pins were surgically attached to the tibia, fibula, talus and calcaneus. Markers that sat at the distal end of the pins tracked skeletal motion. Three skin markers were adhered laterally on each shank, heel and forefoot. Findings gathered revealed a mean maximal difference of less than 5 □ between skin and bone joint rotations. The smallest absolute difference was found for plantar-dorsi flexion.

2.1.1.1 Limitations in the use of intra-cortical pins

Preliminary studies (Levens et al., 1948; Lafortune and Lake, 1991) used only one marker on the distal end of each pin. VICON technology relies on a 3 marker system to define movement on each segment across the x, y and z axes, thus information gathered in these studies would not be considered to be sufficient for gait analysis.

The use of intra-cortical pins to track skeletal movement gives accurate results when used in gait analysis, however for use in a clinical setting it is far from ideal. Limitations of this technique include the surgically invasive procedure along with the impractical execution of the task, in terms of willing volunteers, time constraints and ethical reasoning. Although it was stated (Levens et al., 1948) that subjects did not report feelings of discomfort, it is likely that surgical implementation of the device would have resulted in some kind of mild trauma. In addition, it would be expected that an external fixture to the leg would result in an unnatural walking pattern. It should be noted that whilst preliminary studies looked at a vast selection of volunteers (26), numbers dropped to between 1 and 3 for each case looked at, thus giving limited results.

This procedure is accurate nonetheless, and data gathered here can be used as a standard comparison against other techniques.

2.1.2. Techniques based on external fixators

External fixators are clinically used for the healing of bone fractures. The device relies on a series of strategically placed pins that hold the bone in place. The distal ends of the pins are attached to a fixed frame that sits outside the body.

In 1992, tissue displacement around the knee and hip joints was investigated (Angeloni et al., 1992). An external device was affixed to the tibia or the femur. The rigidity of the frame allowed definition of a set of axes. An array of markers was lined up over the skin following a map of the anatomical landmarks: greater trochanter, lateral epicondyle, head of fibula, lateral malleolus. Markers were also placed over the external frame, covering the entire segment**.** The set up of this experiment can be seen in Figure 2.3.1.

Results of this study can be seen in Table 2.4. The root mean square (RMS) that deviates from the average value of displacement in the x, y, z directions are notably larger on the skin markers at the anatomical landmarks than the markers mounted on the rigid external frame. The same authors published a similar paper in 1996 (Cappozzo et al., 1996). This series of experiments was set up in much the same way as the previous, with markers placed over anatomical landmarks and covering the rigid frame. Subjects were cyclically analysed both in gait and whilst demonstrating elementary movements, such as: cycling on a bike, hip internal-external rotations and flexion of the lower limb. Results from the walking trial can be seen in Table 2.5. These results showed that skin displacement with respect to the underlying bone was considerably high. During movement, skin markers reached displacement of 10-30mm compared to their fixed counterparts, which displayed only a few millimetres of displacement. The authors of this paper also noted that the magnitude of soft tissue artefact seemed to be entirely dependent on the region. The anatomical landmark relative to the joint angle dictated the soft tissue artefact irrespective of the motor task performed. During execution of a hip internal-external rotation of 45 °, there was found to be a 6–28 ° discrepancy. Rounding up these results, it was concluded that inaccuracies in flexion-extension motion, abduction-adduction motion and internal-external rotations can reach values of 10, 20 and 100% respectively.

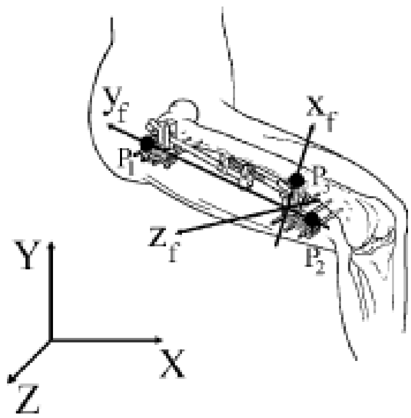


Fig. 2.3.1. Example of a femur fixation device. An external metallic frame is rigidly attached to the bone by means of intracortical bone screws. the three markers (full circles) on the fixator are depicted. The set of axes rigidly associated to the femur (x(f), y(f), z (f)) can then be described in a laboratory frame (X, Y, Z). Adapted from Angeloni et al. (1992)

Table 2.4

Directional displacement of the skin during walking(mm) (greater trochanter: GT, lateral epicondyle: LE, head of the fibula: HF, lateral malleolus: LM) and plate-mounted markers strapped to the thigh (T1, T2, T3, T4) and on the shank (S1, S2, S3, S4).

	X	*Y*	*Z*	*RMS*
Femur				
GT	5	14	8	5
LE	6	16	9	6
T1	7	3	8	3
T2	12	4	9	4
T3	5	5	6	3
T4	4	4	8	3
Tibia				
HF	11	10	7	5
LM	18	13	8	7
S1	8	11	10	6
S2	7	12	6	4
S3	6	7	6	3
S4	9	10	6	4

Table 2.5

Root mean square values of the effect of soft tissue artefact on bone orientation (mm) Three scalar components along the antero-posterior, longitudinal and medio-lateral axes are reported walking task (m6 and m7 are on the shank, LM an HF as above)

Clusters	*Walking*		
m6-m7-LM	1.5	2.5	1.5
m6-m7-HF	1.5	2	2
HF-LM-m7	1	3	3
HF-LM-m6	3.5	2.5	2.5

Adapted from Angeloni et al. (1992).

2.1.2.1. Limitations in the use of external fixators

Comprising similar properties as an intra-cortical pin, an external fixator has direct contact with the bone with the additional advantage of a rigid outward structure that acts as an axis associated with the underlying bone. However, subjects analysed were healing from bone fractures and so a normal gait cycle is unlikely. An external fixator is an incredibly bulky device; it is difficult to imagine how this would not interfere with the subject's walking cycle.

2.1.3. Techniques based on percutaneous trackers

An object that is placed percutaneously simply means it passes through the skin. Percutaneous skeletal trackers descend from a rigid structure and are attached to various parts of the bone. Halo pins, which are attached to a metal ring that encircles the patient's leg, are inserted into the periosteum and an array of reflective markers cover the device. Holden et al. (1997) used percutaneous trackers on the shank of three volunteers. Trackers were attached to the tibia and fibula. A surface mounted target was attached to a shell on the mid-shank in addition; skin markers were strategically placed to cover the foot. This set up can be seen in Figure 2.4. A bone-embedded anatomical reference system that incorporated both the skeletal and skin markers was generated using a static anatomical calibration trial. Soft tissue artefact was calculated using the difference in displacement between percutaneous trackers and skin markers in 3 dimensions. At 8% of the gait cycle, it was reported that there was a 4 ° mean rotation error about the long axis. The highest error reached throughout swing stance was 8 ° in one subject. Looking at the range of motion about the medio-lateral and anterior-posterior axes, rotations were less than 3°. It should be noted that the greatest errors were found to be along and around the shank longitudinal axis. Displacement values of 6.0mm were reached in the transverse plane and as high as 10.5mm longitudinally. Subjects reported

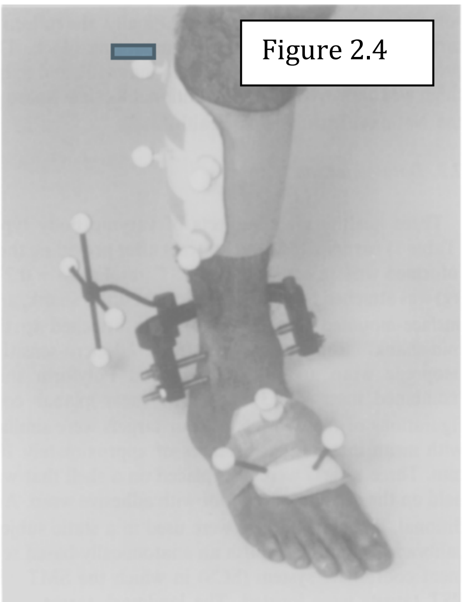


Figure 2.4

little difficulty with walking whilst wearing the percutaneous tracker device. Stride lengths were recorded at 1.2 – 1.4m at speeds of 1.1 – 1.3 m/s. It was found that knee joint forces and moments were prominent during stance phase. A force of 39N was recorded in the shank medio lateral direction and 9Nm about the medio lateral axis. Authors of this paper concluded that soft tissue artefact was dictated by the region and position of the bony segment and the effect it had on joint forces and joint moments.

In another study (Manal et al., 2006), percutaneous trackers were used to determine an optimal surface tracking marker set for tracking motion of the tibia during gait. Comparison of these methods revealed information about how regional aspects affect soft tissue artefact. Using anatomical landmarks to define anatomical frames for the foot, shank and thigh, markers were adopted to track movement. Eleven different marker sets were assessed based on differences in the location of the markers, the method of attachment of the markers and physical construction of the set. A percutaneous skeletal tracker was anchored to the medial and lateral aspects of the foot. Figure 2.5 describes the various methods of placement and construction. The magnitude of the rotational deviation across the entire stance for each marker was quantified by calculating the root mean square difference (RMSD).

$$RMSD_i = \sqrt{\frac{\sum_{f=1}^{frames} RHA^2_{(i,f)}}{frames}}$$ $$RMSD_i = \sqrt{\frac{\sum_{f=1}^{frames} RHA^2_{(i,f)}}{frames}}$$

Where i = surface marker set, f = frame of video data, and "*frames*" representing total stance duration in video frames. Results from this study indicate that the only governing aspect that influenced rotational estimates of the tibia during gait was the location of the marker array. It was found that values of higher skin displacements were found distally. Regions that were examined can be seen in table 2.6. and 2.7. The distal lateral shell underwrapped ranked highest as the best set of markers.

Figure 2.5

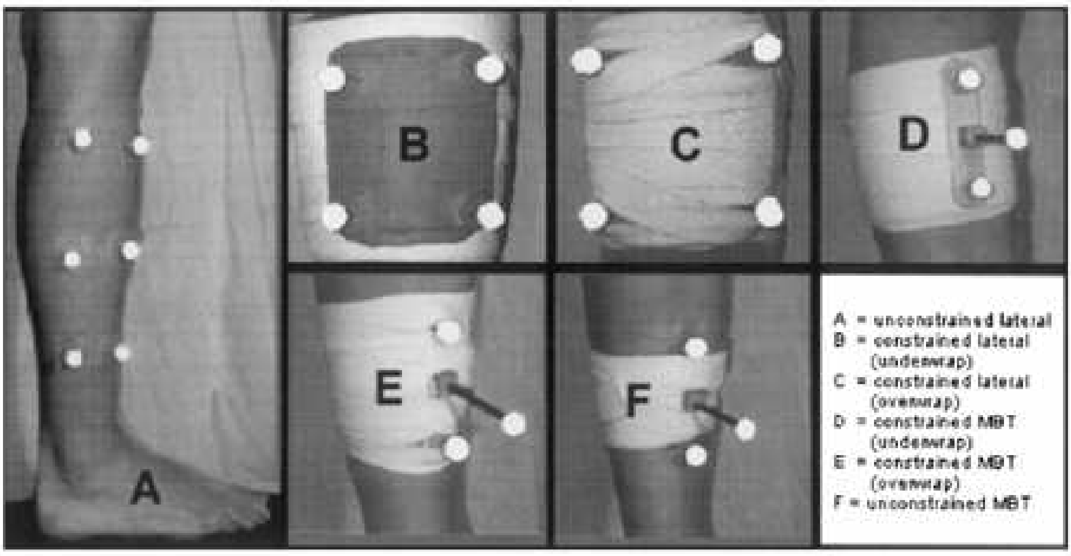


Figure 2.5. Illustration of the various marker sets in the study. Adapted from Menal et al., (2006).

Table 2.6

Root mean square deviations for the factors of location, attachment method and the physical characteristics of marker arrays over the lateral shank

Factor	*Location*		*Attachment method*		*Physical characteristics*	
Condition	*Proximal*	*Distal*	*Underwrap*	*Overwrap*	*Unconstricted*	*Constricted*
Mean (◦)	2	1.7	1.7	1.9	2	1.8
Variance	0.3	0.2	0.1	0.2	0.5	0.1

Table 2.7

Root mean square deviations for the factors of location, attachment method and physical characteristics of marker arrays over the medial border of the tibia

Factor	*Location*		*Attachment method*		*Physical characteristics*	
Condition	*Proximal*	*Distal*	*Underwrap*	*Overwrap*	*Unconstricted*	*Constricted*
Mean (◦)	1.9	1.9	2	1.7	1.9	1.8
Variance	0.2	0.2	0.2	0.2	0.3	0.1

Adapted from Manal et al. (2000)

The percutaneous tracker technique offers a solid reference system by means of the rigid structure that the pins decend from. On top of which there is direct contact between the pin and the bone. Figure 2.4 shows the set up of the experiment.

2.1.3.1. Limitations in the use of percutaneous trackers

The addition of the bandage to hold percutaneous markers in place could restrict walking movement and give erronous values of soft tissue artefact of skin markers as the span of tissue that covers the leg is not moving as it normally would. Subjects reported little difficulty walking, however the attachment of a surgically implemented and bullky device would more than likely hinder gait.

2.1.4. Roentgen Photogrammetry

Roentgen photogrammetry is used as a means of evaluating two dimensional kinematic behaviour of a joint in motion. The use of 2-dimensional Roentgen photogrammetry was used to quantify soft tissue movement around the ankle during inversion/eversion manoeuvres (Maslen and Ackland, 1994). Ten volunteers participated in the study where spherical lead markers were glued onto the skin over the two maleoli, the navicular tuberosity, the sustentaculum tali and the base of the fifth metacarsal. Data was gathered via lateral view radiographs. Results reflected a varying degree of soft tissue movement around separate regions. Mean displacement between skin markers and underlying bone ranged from 2.7-14.9mm. Large magnitudes of skin displacement were found under the two malleoli markers.

In a later study composed by Tranberg and Karlsson (Tranberg and Karlsson, 1998) a foot in the neutral 20° of dorsi flexion and 30° of ankle plantar flexion was analysed using Roentgen technology. Data gathered from 6 volunteers showed a soft tissue displacement of up to 4.3mm. A similar pattern that arose from the previous experiment was found for large magnitudes over the two malleolus. However a comparative difference between the two studies arose from the location of the largest movements. In this study greater tissue displacements were found to be proximally instead of distally. Markers mounted on the foot recorded a movements ranging from 1.8 to 4.3mm that corresponded to the underlying bones.

The concept of using x-ray fluoroscopy along with a mathematical model (Sati et al., 1996) was used to quantify relative skin movement to bone. An array of 3mm stainless steel spherical markers were taped on to the medial and lateral aspect of the condyles, and on the lateral aspect of the thigh, the set up can be seen in picture one. Using x-rays medical imagery, the silhouette of the skeletal movement is monitored and measurements between this and the stainless steel makers were made. The positioned markers were defined with respect to an orthogonal set of axis fixed with respect to bone. Three volunteers were asked to move their foot backwards in a sweeping motion so as to hit their buttock. This motion at the knee imitates one made at 65°during the swing phase of gait and lets the movement be caught in a fluoroscopic window. Data gathered from this experiment can be seen in Table 2.8. The results showed that marker movement ranged from 2.5mm RMS to 17mm RMS laterally. Along the antero-posterior and vertical direction, peak-to-peak root mean square values of 42.5 and 20.6mm RMS were found. Looking at the medial aspect, the root mean square skin movements were contained in the range of 2.1 to 17.1mm and peak-to-peak values were found to be 31.0 and 39.2mm. It was noted that marker location dictated the varying magnitudes of soft tissue artefact, with largest displacement recorded near the joint line.

Table 2.8

	RMS lateral marker movement on subjects (mm)								
	Sub. 1			*Sub. 2*			*Sub. 3*		
Marker no.	*RMSd*	*x*	*z*	*RMSd*	*x*	*z*	*RMSd*	*x*	*z*
1	8.9	23.57	3.19	16.59	41.6	12.21	16.77	42.51	20.6
2	9.13	21.53	12.92	13.85	31.22	19.74	15.4	21.33	33.27
3	8.02	17.07	12.92	12.89	22.34	24.73	12.57	33.19	14.35
4	3.47	6.15	7.8	7.61	18.21	9.37	7.78	21.37	6.13
5	2.93	6.92	4.45	9.01	16.44	18.37	6.51	13.79	11.2
6	8.72	9.9	23.31	5.86	12.31	12.63	7.18	8.02	19.08
7	6.57	6.23	15.09	5.87	10.63	12.28	6.03	8.24	14.93
8	2.58	3.19	6.31	9.45	6.89	12.1	4.96	6.13	13.2
9							2.47	4.15	6.19
10							2.86	3.86	7.73

	RMS medial marker movement on subjects (mm)								
	Sub. 1			*Sub. 2*			*Sub. 3*		
Marker no.	*RMSd*	*x*	*z*	*RMSd*	*x*	*z*	*RMSd*	*x*	*z*
1	4.7	12.2	6.8	14.95	38.83	16.41	15.4	43.8	4.95
2	5.03	11.78	8.11	17.12	31.03	39.21	14.13	38.13	13.6
3	3.52	2.33	10.03	7.8	20.02	7.34	11.5	24.34	22.01
4	2.15	3.72	4.94	8.41	17.84	16.99	8.37	20.78	15.36
5	5.53	4.01	15.47	10.83	12.01	24.34	8.2	23.7	2.53
6	3.26	6.26	7.57				8.81	22.65	13.95
7							5.85	15.56	7.38
8							5.24	16.42	1.21
9							8.54	5.95	24
10							3.57	8.91	6.82

Adapted from Sati et al. (1996)

In 2002, a conference paper was submitted (Tashman and Anderst) that described impact movement analysis. A pioneering study in this area, research was carried out by looking at one- legged forward hopping. Two markers over the medial and lateral epicondyles were tracked. Patients of total knee surgery who had had tantalum beads implanted at the time of knee surgery were studied here. The reference system was constructed from femur and tibia motion. Peak-to-peak values in the range of 5-31mm were recorded after foot impact. It was found that the subject, marker and direction were determining factors for the time of impact to peak displacement, dominant frequency and the magnitude of the transient component of displacement.

Stereophotogrammetry and 3-dimensional fluoroscopy were applied in a combined effort to categorize soft tissue artefact. A patient, who had received a total knee replacement, performed sit-to-stand and stair climbing motor tasks. Nineteen skin markers were placed on the thigh and ten on the shank. Synchronisation was obtained by a reflective radioplaque marker that was adhered to the patella and spacial registration was got by three reflective radioplaque markers placed in the fluoroscope field of view. Following the regional patterns that have emerged from studies on soft tissue artefact, markers on the thigh exhibited greater displacements than those on the shank. Values for soft tissue artefact reached 40.0, 51.5 and 55.3 mm along the respective antero/posterior, medio/lateral and vertical directions. Rotations around the tibio/femoral aspects revealed a root mean square of error ranging from 250% distally to 360% proximally for sit-to-stand analysis. Stair climbing revealed a 135% distal and 185% proximal root mean square rotational error.

2.1.4.1. Limitations of roentgen photogrammetry

Roentgen photogrammetry uses radiographs to obtain x-ray images of lead skin markers as they move relative to bone. An obvious problem with this technique would be the health dangers associated with exposure to radiation. Additionally, in the study looked at (Tashman and Anderst, 2002), the subject had undergone knee surgery. Thus movements of this subject may not correspond to the movement of a subject who had not undergone surgery. Furthermore, this technique can only observe motor-tasks performed within a small window due to the set up of the radiograph. Comprehensive gait analysis is not possible.

2.1.5. Use of Ultrasonic Technology

Ultrasonic technology is widely used in medicine. Waves penetrate a medium and measure the reflection signature or supply focused energy. Typically used in somography to build pictures of the foetus in the womb, it has also been used for studies investigating soft tissue artefact.

In 2005, radiographic images of the thoracic region of the spine (Yang et al.) were taken at the extremities of motion (extension and flexion). Skin markers were applied to strategically selected landmarks along the vertebrae (between T7 and S1). Quantifiying results, it was found that soft tissue artefact in this region was 4.23 ± 33.59mm. Such a large variance could be down to too low a sampling rate.

Soft tissue movement around the thoracic axial rotation and single arm elevation was investigated in a study led by Heneghan and Balanos (2010). Skin markers were placed at intervals along the vertebrae at C7, T5 and T11 and subjects were asked to move from a neutral position to one where the arms were folded across the chest. An ultrasound was acquired at both positions. Motion was measured using a combination of ultrasound imaging with motion analysis of the ultrasound transducer (Heneghan, 2009). Results showed that high values of soft tissue artefact could be attributed to the type of activity. The mid thoracic region (T6), exhibited soft tissue artefact of 16.57 RMSmm.

The studies above describe a non-invasive, readily accessible method of distinguishing soft tissue artefact in a clinical setting. Adaption of this technique to address soft tissue artefact in the lower limb would be a possible area of interest for future study.

2.1.5.1. Limitations in the use of ultrasonic technology

The type of movement that could be analysed using this technique would be limited due to the small area in which movements would be performed.

Standard flexion-extension of the knee joint could be examined using ultrasound, but full gait may not be possible.

2.1.6. The Use of Accelerometers

An accelerometer is a type of transducer that measures inertia forces. Dynamic forces that can be determined by an accelerometer include movement and vibration.

In 1978, J.E. Smeathers reviewed transient vibrations , shock absorptions and transmissibility of vibrations caused by heel strike using skin attached accelerometers. Vibrations were measured by considering variances in an oscillating wave that was obtained using an accelerometer. Vibration is said to be transient when an oscillation dies down due to damping. The damping, in this case, is down to the properties of the soft tissue that surrounds the bone. Smeathers investigated the shear stiffness of the soft tissue by performing a nudge test. An accelerometer was attached to the skin using tape and the nearby skin is displaced, or nudged and quickly released. From this the damping factor, damping frequency and natural frequency can be extracted. Results from this revealed that the damped frequency varies between 10 and 20Hz. Damping factor lies between 0.25 and 0.5.

A more recent study looked at the motion artefact of skin mounted accelerometers under different attachment conditions (Forner – Cordero et al., 2008). Three sets of experiments were performed. An adequate excitation stimulus was selected after evaluating different excitation procedures. The selected stimulus was then applied under different attachment conditions. And lastly, a third experiment was used to test the model. The excitation method chosen was a heel drop movement due to its ability to discriminate between different attachment conditons and it showed lower variability. Displacement of the accelerometer at heel drop and gait impact were evaluted under unloaded, loaded and load plus mass conditions. The unloaded condition described the accelerometer being stuck to the skin. Loaded described the use of an elastic bandage to compress the skin. And the load plus mass was an elastic bandage with added mass. A resultant output voltage was obtained from the reading of the accelerometer. From this the natural frequency was collected. Results revealed that the manual displacement was not a valid testing method, this was seen by application of MANOVA statistics that showed an abnormal frequency reduction with increasing spring of stiffness. From the second experiment, authors concluded it was possible to use an accelerometer if the bandage is strapped firmly around the limb.

2.1.6.1. Limitations in the use of accelerometers

Limitations of this study could include the low number of subjects invloved. Differences to the level of pressure due to bandaging that each participant could torelate was also an uncontrolable variable. The simple model for determining damping coefficients from the nudge test may not be valid to represent the soft tissue motion in three directions. It should be noted that noise and any electrode potential given by the muscles were not subtracted from the accelerometer readings and that walking frequencies were assumed to be between 4 and 6Hz, and this was not tested here but taken from referenced studies.

2.1.7. Development of a string type flexible displacement sensor

A recent study (Dohta et al., 2008) described the development of a flexible displacement sensor that could be applied to measure skin displacement. Developed for use in nursing care for the elderly, the skin displacement sensor is a byproduct of the flexible displacement sensor. The flexible displacement sensor consists of two fixed electrodes, slide electrodes and a nylon string coated with carbon. The skin displacement sensor is pasted onto the skin using a plaster. When the body and skin moves, the distance between the two sensor changes. The change causes pushing and pulling forces that act on one end of the flexible displacement sensor. This method measures the displacement of the skin. This was tested on the arm of one subject, and it was placed over the bicep muscle while the subject performed flexion tasks. The sensor was placed in two directions to aquire movement in the x and y directions. This method was used as an indirect way of measuring body movement, by monitoring skin displacement with body angles.

This is a novice system that effectively measures displacement over the different regions of the skin. The noninvasive approach is ideal for clinical assesment.

2.1.7.1. Limitations of a string type flexible displacement sensor

While this could be applied to gait analysis, it is uni-axial which is limiting. A tri-axial measurement system would be more desirable. Another limitation is the large size, this one measures at 100mm.

2.2. Misrepresentation of Anatomical Landmarks

The misrepresentation of anatomical landmarks causes major problems in gait analysis. It is the representation of anatomical landmarks that provides an axes frame for the VICON to work within. Locating of anatomical landmarks is expected to be precise, however every person exhibits different bone dimensions. The properties of soft tissue that cover the bone means that errors pin pointing specific locations can occur. Attempts to locate anatomical landmarks have been made, these include; the solidification procedure; multiple anatomical landmark calibration; pliant surface modelling; dynamic calibration; point cluster technique and global optimization. The techniques explored reveals a progression in treating the limb from the traditional rigid body structure to the more apt non-rigid body structure.

2.2.1. Solidification Procedure

In 1995, Ch'eze et al. developed a technique called the "solidification procedure" that attempted to define skin marker trajectories in accordance with a rigid-body segment . This technique works by distinguishing the 3-marker triangle that deforms the least during a single motion. This gives a defined, time varying triangle, the solid shape of this is calculated using the mean of each angle. This along with the single value decomposition algorithm, which is the factorization of a matrix in mathematics, is used to determine a best fit solid to each measured triangle. Visualisation of the simulated marker coordinates can be seen in Figure 2.2.1. This technique was used to identify kinematic errors experienced during gait. One subject was analysed with rigid clusters of markers adhered to the thigh and shank. Perturbed marker trajectories were obtained by the introduction of artificial noise that represented typical tissue displacement during gait. Kinematic errors around the knee were observed. The results communicate that this procedure gives a similar reduction in kinetic error as the least squares method. The authors concluded that the main advantage of this procedure is that it can identify erroneous frames and it allows incorporation of the rigid body theory, which is easy to establish.

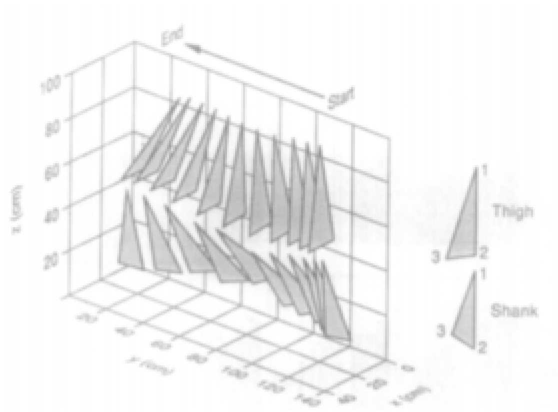


Figure 2.2.1. Visualization of the simulated marker coordinates used in the numerical experiments. The data corresponds to the unperturbed movement of the shank and thigh segments during the swing phase of gait. Markers on the thigh correspond to (1) greater trochanter, (2) the lateral femoral epicondyle, and (3) the medial femoral epicondyle, while markers on the shank correspond to (1) head of the fibula, (2) the lateral malleolus, and (3) the medial malleolus.

Adapted from Ch'eze et al. (1995)

2.2.1.1. Limitations of the solidification procedure

Although the rigid body model is easy to establish, a feature that would be of much use in a clinical setting, it projects a misguided representation of the limb in motion. Selecting a triangle that deforms the least would not be effective if the triangle moved uniformly in one direction. A triangle could have kept it’s structure but migrated considerably during motion, which does not determine skin movement relative to the bone. The introduction of artificial noise to represent typical displacement could also be a flawed approach. Soft tissue artefact is not a random occurrence, it is a response to particular movements and the characteristic properties of the tissue. Both studies only investigate the results from one subject, the findings here could be specific to one person.

2.2.2. Multiple anatomical landmarks calibration

Cappozzo et al. (1995) details the Calibrated Anatomical Landmark System. This involves one static calibration of selected anatomical landmarks to locate a clustered technical frame and build an anatomical frame. Anatomical landmarks are identified by means of a pointer that mounts two markers. An improved technique (Cappello at al., 1997) incorporated a double calibration between the anatomical landmarks and the central technical cluster. A single static calibration was taken at flexion and another at extension. This technique moves away from the rigid body theory as it captures two coordinates at the terminal points of joint motion. Cappozzo et al. obtained data (Cappozzo, 1996) that supported the idea that during motion, the coordinates of the anatomical landmarks, in the relevant frame of reference, constantly change. Thus the tangent between two calibrations can be isolated and used to model motion.

An experiment was carried out where one subject performed a cycling exercise wearing a femoral external fixator. The greater trochanter, lateral epicondyle and the medial epicondyle were calibrated at maximum extension, which was the pedal down position and again at the minimal flexion, pedal up position. Using the standard Standard Deviation Technique, (Ch'eze et al., 1995) an optimal tangent between the two extremes of cycling motion was estimated. An established conversion, or transform was then applied to selected anatomical landmarks. Cluster shape variation and anatomical displacements were obtained by looking at cluster and anatomical landmarks in the maximum extension and following these through to minimal flexion. Assuming time was an independent variable and including calibration parameters, a time-varying model was produced. Results showed that by applying a multiple calibration and utilising the time varying model, lower values of error were obtained. The root mean square error recorded at a trajectory spanning from the greater trochanter was reduced by 5mm. Whilst it was found that root mean square error values at the femur dropped by 1 □ in orientation 3.5mm in position. Authors concluded that a multiple calibration technique produced encouraging results, although an enhanced procedure that could characterize the sliding and displacement of skin during execution of motor tasks should be considered.

2.2.2.1. Limitations of the Multiple Landmark Calibration

Multiple landmark calibration is a good approach as it recognises that the co-ordinates of anatomical landmarks change during motion. However it does not tackle problems that arise from soft tissue over the anatomical landmarks when initial placement of the markers is being applied. There is only one subject investigated in this study, this could result in biased findings. The use of an external fixator creates resistance on the tissue when normally there would be none, it also implies that the subject is recovering from an injury and so motion and forces applied could be slightly impaired.

2.2.3. Pliant Surface Modelling

Degree of freedom, in mechanics, represents the number of independent displacements or rotations that a body or system has undergone in space (Soderkvist and Wedin, 1993). In 1998, Ball et al. modelled the pliant surfaces of segments of the leg during gait by comparing mathematical models of a rigid body structure to that of a non-rigid one. Generally speaking, for a rigid body the degree of freedom is governed by $d(d+1)/2$, where d is the dimensions in space (Soderkvist and Wedin, 1993) . The traditional rigid body model works within 6 degrees of freedom and is based on the assumption that the limb segment and surface is rigid and stays constant. The newer, non-rigid body model works within 12 degrees of freedom and uses rigid rotations and translations, adding on values that represent the shearing and displacement changes that accompany skin stretching, muscle movement and general inertia. This was modelled by using an algorithm for computer graphics that can modify the shape of a virtual object. An affine deformation matrix expressed scaling, shearing, rotation and translation.

A standard comparison was established by using intra cortical pins to track the movements of three subjects. Subjects performed walking tasks on a treadmill at three different speeds. The rigid body technique produced errors of 4.8 and 3.8mm when reconstructing real femur and tibia segment position. The non-rigid body model saw femur error reduced by 45% and tibia segment error by 56%. Authors noted that between the two methods there was no considerable difference in segment pose for different walking speeds.

2.2.3.1. Limitations of Pliant Surface modelling

Using intracortical pins as a baseline is an accurate and verified method of tracking skeletal movements. A clear advantage to using a non-rigid body model compared to a rigid body model when addressing soft tissue artefact is that it reflects the true properties of the limb. It also produced an encouraging reduction in error. Yet this method still yields 2.64 and 1.672mm error in the femur and tibia segment, respectively. There are questions of validity that arise from the computer graphic software that determine the properties of the tissue that covers the bone of the limb. The generalised approach to quantify soft tissue should be reassessed, a prominent feature across the studies examined is that soft tissue artefact varies for each individual.

2.2.4. Dynamic Calibration

In 1999, Luccetti at al. developed a technique to quantify soft tissue artefact in the estimation of knee joint kinematics. The relationship between cluster technical frames and anatomical landmarks determined a dynamic model that was both subject and task specific. A method of least squares was used to find the position and orientation of bones during motor tasks. A pelvis brace was covered with 4 markers. Another 4 skin markers were placed on the shank and 5 on the thigh. Locations of the thigh, shank and the pelvis acted as co-ordinates to determine a standard cluster technical frame. Another was determined from the thigh and segment only. Data was collected from two adult male volunteers standing upright, and then again during gait. Subjects were then asked to perform several motor tasks that included hip flexion/extension followed by abduction-adduction; a lower limb pedulum swing; and a simulation of the swing phase of walking, that was achieved by rotations of the hip and pelvis. Anatomical landmarks were calibrated and the functional approach was used to estimate the hip joint centre. This functional approach is based on an examination of the kinematics of various postures to define a model that covers cyclical movement, and the type and nature of the movement when estimating hip joint centre (Begon et al., 2007) A rigid body model approach was used to estimate positions of the medial and lateral epicondyles. Markers on the shank were used here, as previous studies (Sati et al., 1996) indicate that soft tissue moves less during motion on the shank than the thigh. Subjects were asked to stand straight with the knee locked, the locations of the markers acted as a reference point and were used to identify anatomical landmarks and their relation to hip flexion-extension, abduction-adduction and inversion-exersion. These results made up the "table of artefact". Estimative values of hip rotations were acquired by initially selecting the smallest distance between the table of artefact and location of skin markers that was captured during gait analysis. Revised anatomical landmarks were calculated by subtracting the "artefact value" from those collected at the final position. Building a new anatomical frame from the revised anatomical landmarks allows knee translations and rotations to be computed.

This paper also details a second, more sophistocated approach to this problem by analysing a patient who wore a one degree of freedom knee arthoprosthesis. Comparitive results between a rigid model (without compensation for artefact) and a non-rigid model (with compensation for artefact) can be seen in Table 2.9**.** The rigid body model resulted in knee joint translations and rotations that exhibited a root mean square error of up to 14mm and 6˚ respectively. A non-rigid body approach gave more favourable results of 4mm and 3˚ respectively. A reduction in error was found for femur and tibia poses using this technique against the method of least squares. A reduction of 4mm was found for knee joint translations and 3 □ for rotations. This technique shows less soft tissue artefact at the hip than before. Results from knee kinematics of two subjects revealed that there was less distinction in soft tissue artefact when compensation measures are applied.

Table 2.9

Root mean square errors (mm)associated with estimates of knee joint kinematic variables with and without compensation

Subject (motor task)	*Estimate*	*a*	*b*	*c*
LL (limb swing)	Without comp.	3.9	10.9	4.9
	With comp.	1.9	2.1	1.4
GC (limb swing)	Without comp.	5.8	5.8	3.3
	With comp.	2.8	2.8	1.6
TM (walking)	Without comp.	10.1	13.2	4.2
	With comp.	2.5	1.7	0.8
TM (walking)	Without comp.	13	11.9	6.7
	With comp.	2.9	3.6	4.5

Root mean square values (mm) of the differences between estimates of knee joint kinematic variable obtained with two different sets of thigh markers during walking trials

Subject (walking speed)	*Estimate*	*a*	*b*	*c*
LL (1.20 m/s)	Without comp	*19.6*	*12.1*	*9.2*
	With comp	*5.1*	*4.2*	*5.8*
GC (1.12 m/s)	Without comp	*11.6*	*18.4*	*5.7*
	With comp	*2.6*	*7.7*	*3.8*

In both cases, a is mediolateral displacement; b is antero-posterior displacement; c is distraction. Adapted from Sati et al. (1996)

This technique addresses and overcomes some of the fundamental flaws that has been seen in previous papers when calibrating. Calibrating anatomical landmarks during motion compensates for rotations and translations that alter the location of the anatomical landmark. Revising the anatomical frame in accordance with the constantly changing anatomical landmarks is a thought out and well presented solution to reduce errors associated with soft tissue artefact.

2.2.4.1. Limitations of Dynamic Calibration

Limitations of this technique could be that gait itself is not analysed, rather simulated movement that copy gait when the subject is standing still. This is due to parameters posed by the set up of the experiment. Another limitation could be that tracking movement using the single degree of freedom prosthesis, although does generate a complete axis, mimics movement that is not representative of a normal leg during gait. The number of subjects examined in this study could be too few to produce information over a broad enough range.

2.2.5. Point cluster technique

Andriacchi et al. (1998) investigated the artefact associated with non-rigid body movement associated with selected points from skin markers. The method involves placing a cluster of markers so as to cover the limb. Each point is assigned an arbitrary mass. Calculations determine the centre of mass and inertia tensor of the cluster points. A coordinates system was established using the eigenvalues and eigenvectors of the inertia tensor. The aim of this technique is to minimise eigenvalues by altering the mass of each marker at each step. If the segment is rigid, then the eigenvalues do not vary, and so using this as a point of reference, variations for non-rigid body were determined. Systematic and random errors were introduced into a fixed cluster of points to test a simulation model. The simulation saw a substantial reduction in error due to non-rigid body movement. Applying this method to 10 volunteers who were analysed in gait, the authors found that knee rotations and translations obtained from the point cluster method produced similar results in line with those gathered using intra-cortical bone pins (Karlsson and Lundberg, 1994).

In a further study, the same authors extended the transformation equations to the general deformation case (Andriacchi and Alexande, 2007). An eight marker cluster set was used during 50 simulated trials and on one patient who had an external fixator device attached to the shank. The external fixator offered a rigid axes for the system, this single test saw a reduction in the error obtained for overall pose. Differences between internal deformation technique and rigid body model are highlighted by changes in location and orientation. Figure 2.911 the displacement and orientation with respect to time of the rigid body model and internal deformation technique correction. From these results it can be seen that location error yielded a reduction of 17mm and orientation error yielded that of 29mm per time step. It was stated that skin motion could have been restricted by the number of pins that transceneds the skin (Alexander and Andricchi, 2001).

Figure 2.9.1

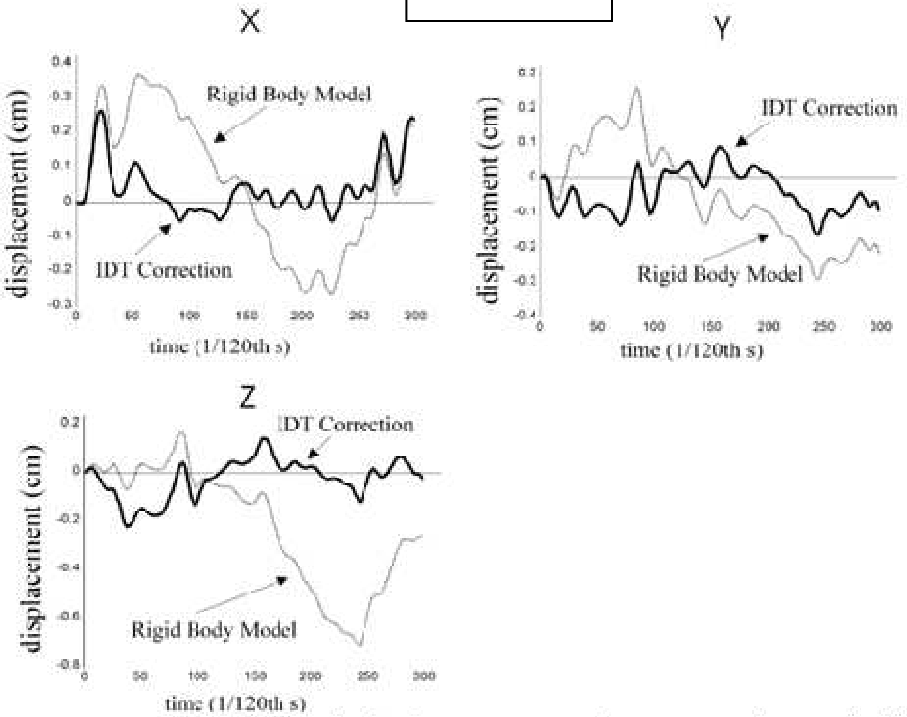


Figure 2.9.1. A comparison for the displacement error relative to a true bone embedded system for a rigid body model and for the inteval deformation technique. Adapted from Andriacchi and Alexander (2001)

<u>2.2.5.1. Limitations of the Point Cluster Technique</u>

It should be noted, however, that both techniques demand a large number of markers to be used, one constriction of additional markers is, that if placed too close together, the VICON can misinterpret a few as one – thus giving false readings. A further drawback is that application of these techniques are time consuming. Authors of this paper concluded that erroneous results could surface from the limited information the skin deformation models contain. That perhaps they cannot cope with the displacement of the markers with respect to the underlying bone.

Tasham et al. (Tasham et al., 2002) was sceptical about the validity of using skin markers to track vigorous movement or extreme impact. Stagni et al. (Stagni et al., 2003) looked at this technique and critisied the dependency that estimated skeletal kinematics have on the modelling form, it may not give accurate representation of the bone frame. It was also noted that estimation of orientation was slight compared to bone position.

2.2.6. Global Optimisation

Global optimisation realises that each segment does not follow isolated movements, and that artefact over one segment will transcend and effect another. Kepple et al. (Kepple et al., 1994) describes an optimisation technique wherein by the position and orientations of the musculoskeletal model is determined using marker coordinates. It was hypothesised that if a more accurate representation of joint motion could be realised, then the estimation of soft tissue artefact could be more precise. This technique involved producing a “weighting matrix” that was calculated by taking the distance between simulated and model-determined marker positions, squaring this value and taking the minimum weighted sum. This method assumes a constant value of soft tissue artefact exists for each segment, however it can distinguish the differences between each segment. Each segment is labelled with its own “weighting factor”. Incorporating artificial noise (Kepple et al., 1994) to the value of each marker, Lu and O’Connor (Lu and O’Connor, 1999) trialled this technique on 20 subjects. Results show the stark contrast between non-optimized technique, that show hip dislocations at 3.88cm and knee dislocations at 3.24cm. Optimised values gathered were 1.33 and 0.69cm respectively. A notable reduction in error recorded for joint motion was found via application of this technique. Authors also noted that the incorporation of a weighted matrix gives a finer tuning of joint location and aids in minimizing soft tissue artefact throughout motion.

The Helen Hayes gait model is a conventional marker set that involves the thigh and shank being traced with the aid of a 10cm wand that captures motion in 3-dimensions. Global optimization was applied to the Helen Hayes gait model (Roren and Tate, 2002). One study, (Charlton et al., 2002) describes a tested system that challenged the repeatability aspect of this technique. A single subject was asked to perform 100 gait cycles, that were observed by 3 physiotherapists. Results reveled smaller recorded distances between the skin markers during walking motion. Bone segment dimensions were also found to be lower using this technique, against the conventional marker model.

In 2003, Cerveri et al. minimized the distances between marker projections obtained during gait and a back projection markers in a 3-dimensional model. A computed algorithm was developed that included a differential model to represent the non-linear behaviour of the motions captured. This allowed accurate estimations of joint kinematics. In a later study, the same authors (Cerveri et al., 2003) described a robust recovery of human motion from videos using kalman filters and virtual humans.

This technique understands that tissue is not disjointed by segments, and attempts to incorporate the rotations and translations of one segment as it transcends another. A wide range of subjects were examined in these experiments, which produced encouraging results.

2.2.6.1. Limitations of Global Optimization

A failing of this technique could be the assumption that each segments posseses a uniform artefact. The differences in soft tissue artefact that can occur at the posterior region of the shank to the prominent bony anterior region that covers the tibia would considered as a descrepancy.

2.3. Summary

A review of the different methods of assessing soft tissue artefact reveals and misrepresentation of anatomical landmarks reveal, in some cases, very different values. The large margins that are documented through out the literature are likely to be caused by the different methods being used to quantify the errors caused by skin markers. Despite varying results, a distinct pattern has emerged from studies. Similarities include: that soft tissue artefact around joints is higher than those over the middle of limb segments; soft tissue artefact is task-dependent; body mass index dictates a variance in soft tissue artefact; soft tissue artefact is prone to both systematic and random errors; soft tissue over the thigh is likely to endure greater artefact than that over the shank. Studies inferred that whilst skin markers could be used for joint flexion-extension, a sceptical eye should be applied to results got from abduction-adduction and internal-external rotations. Some of these techniques also hinder normal gait. It should be noted that capture of normal gait could be compromised when a person is asked to "walk naturally", the deliberate movements and thought process involved could mean that the person is not walking naturally at all.

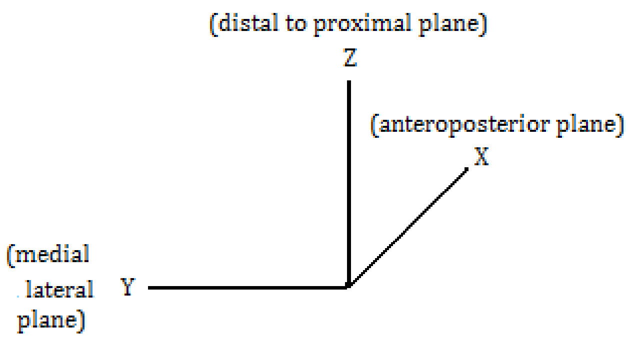
CHAPTER 3. Methodology

Six healthy adult volunteers (4 female and 2 male) participated in an ethically approved investigation that looked at soft tissue artefact using a point cluster technique and values obtained from accelerometers.

3.1. Cluster technique

A 9 camera VICON system was used to capture the subjects in gait using a template designed specifically for this experiment. Subjets were asked to wear a latex suit and reflective spherical markers (13mm in diameter) were attached down the left and right sides of the body. Hypoalergenic tape was used in to attach skin markers at the asis, the lateral thigh, the tibia tubercle, and lateral shank. A 4 point marker cluster was attached firmly with masking tape onto the shank and thigh of each leg. This can be seen in Figure 5.1. The skin markers are used to determine movement in the x, y, and z directions relative to the cluster. The surrounding video graphic system records data in 3-dimensions and generates trajectories based on the movement tracked by the markers. The system was calibrated using static calibration.

The axes in VICON are as follows: the x direction runs along the direction of walking on the floor, data gathered here represents movement in the anteroposterior plane. The z axis is vertical and data captured here is in the distal to proximal plane. The y axis runs along the width of the floor and data captured here lies in the medial- lateral axis.



It was assumed that the clusters did not move relative to the skin.

It should be noted that markings on the skin were made with a pen at each point where a skin marker was placed, this was to ensure that accelerometers being used in the second half of this experiment captured tissue displacement at the same point.

Figure 3.1.

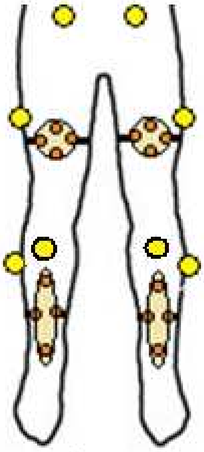


Figure 3.1. Clusters were placed approx half way down the thigh and shank. Skin markers were placed on the tibial tubercle, shank markers were placed approx. 50mm from the shank cluster in the lateral direction. Thigh clusters were placed approx. 65mm from the thigh cluster in the lateral direction.

Each subject was asked to perform 3 cycles of gait, walking on the spot and heel drop. Results were analysed in the sagittal plane with the x axis running along the length of the floor, the y axis perpendicular to this running along the width of the floor, and the z axis shooting vertically from the ground upwards. For the skin marker at the tibia tubercle, the corresponding axes that were used to work out displacement from the shank cluster can be seen in Figure 3.1.1. skin markers at the lateral shank and thigh and the corresponding axes can be seen in Figure 3.1.2. Trajectories of movement were sampled at a rate of 100 Hz. Data was exported to excel files for computational and statistical analysis.

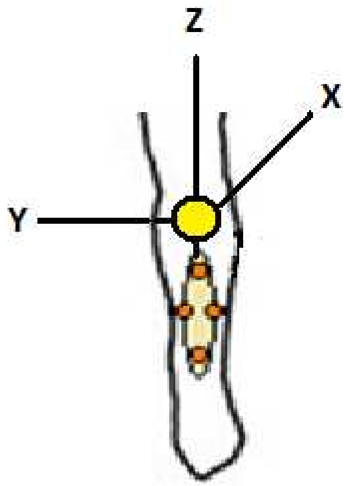


Figure 3.1.1. The marker at the tibial tubercle and the axes from the front view of the tibia.

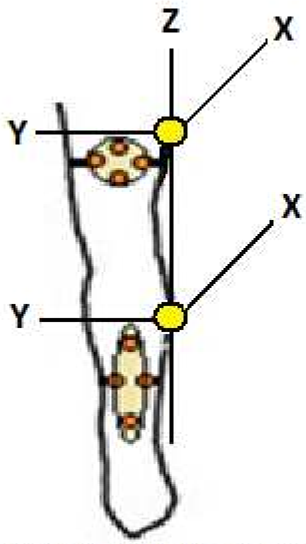


Figure 3.1.2. The marker at the lateral shank and thigh and the corresponding axes from the front view of the tibia

3.1.1. Analysis

Quantification of soft tissue movement was calculated by taking the distance between a skin marker and a one of the markers on the cluster and assessing how it varied during movement. Variances in soft tissue artefact at the tibia and shank were established by markers on the shank cluster. Variances at the thigh were established from markers on the thigh cluster.

3.2. Accelerometers

Three dual-axis accelerometers (Analogue Devices: ADXL32) were attached to a 5V power supply. Once attached to a backing board, each device measured 7 by 10mm and weighed 6g. Each accelerometer was wired with 3m wire to allow the subjects to move without restriction. Acceleration was measured as a voltage output that was captured using Labview software and a DAQ 61125 box.

Each accelerometer had a separate output for the x and y axis. A subtraction of 2.5V was applied at the generated signal to bring the voltage to a baseline.

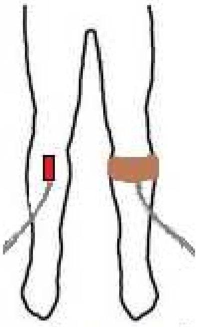
In position on the skin, the x axis of the accelerometer ran vertically up the leg, while the y axis ran laterally along it. Because readings were only taken with the participants upright, there was no need to translate the axis when drawing comparisons between these values and those gathered from the cluster technique. The x axis, in this case captures movement in the distal to proximal plane and the y axis captures movement in the medial lateral plane.

Each participant and the surrounding environment was monitored for frequencies generated by the electrode potential from the muscles and noise, respectively. Using hypoallergenic tape, an accelerometer was attached to the proximal tibia, lateral shank and lateral thigh of the left leg. Each subject was asked to stand still while a reading was made.

A series of nudge tests were performed at the shank, thigh and tibia. This involved applying a sweeping motion with a finger to directly underneath where the accelerometer was attached. Care and attention was made to use the same pressure and sweeping speed each time.

A heel drop test was also performed. Each participant raised themselves up onto their tip-toes and then dropped down onto the heel. The three accelerometers were attached to the skin of the thigh, shank and tibia and data was collected simultaneously.

Figure 3.2.



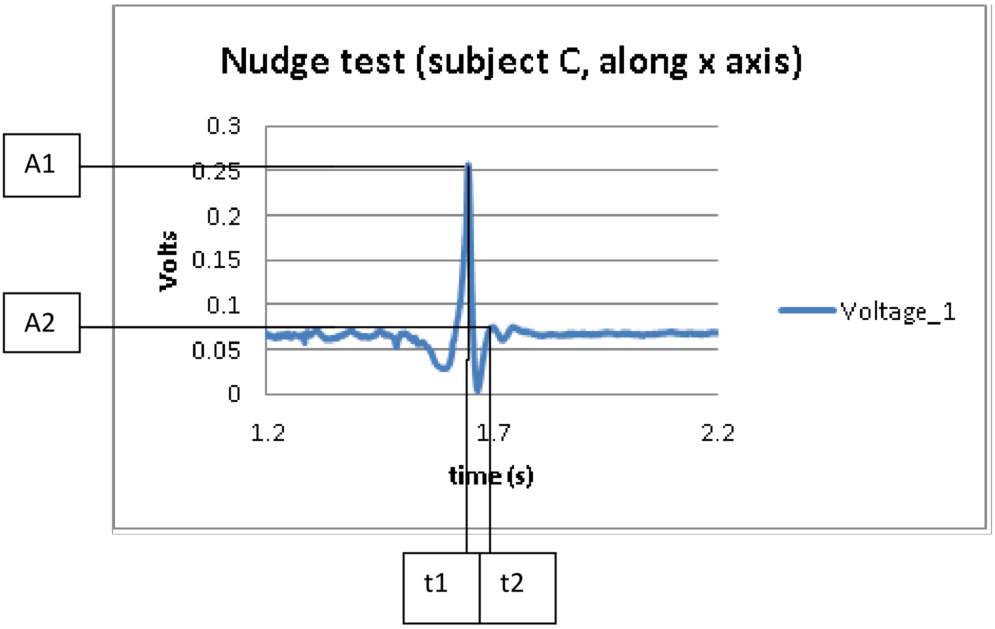
Examining soft tissue movement during the walking trial was done by attaching an accelerometer to the tibia at the right and left leg. The accelerometer on the left tibia was attached using hypoallergenic tape only, and the right accelerometer was attached using tape, and a firmly tied inelastic bandage to hold to accelerometer close to the skin and minimise any artefact that could occur during the trial. This is portrayed in Figure 3.2. Participants were asked to walk on the spot at a natural cadence. The output voltage of the acceleration from each accelerometer was captured using Labview. This method was repeated at the lateral shank and thigh for each subject. Data was exported to excel for analysis.

3.2.1. Analysis

The baseline noise was subtracted from the reading of each participant. The baseline noise was obtained by taking a static reading of the surrounding noise and muscular viration of each participant at the thigh, shank and tibia. It should be noted that to minimise functional errors that could exist between accelerometers, data gathered from one accelerometer in one axis statically was only ever subtracted from data gathered at the same accelerometer and axis during dynamic trials.

To categorize the properties of tissue at each different point, the natural frequency, damping frequency and damping factor gives an indication of the quantity of soft tissue present. These values were calculated from data gathered during the nudge test and heel drop for the tibia, shank and thigh. Values of A1 and A2 along with t1 and t2 were obtained from the peak oscillations of time-voltage graphs. This can be seen in Graph 3.1.

Graph 3.1



The underdamped behaviour of this graph can be expressed as:

Where *de* is the underdamped decrement. This is related to the damping factor by

Rearranging this equation, to work out the damping factor gives:

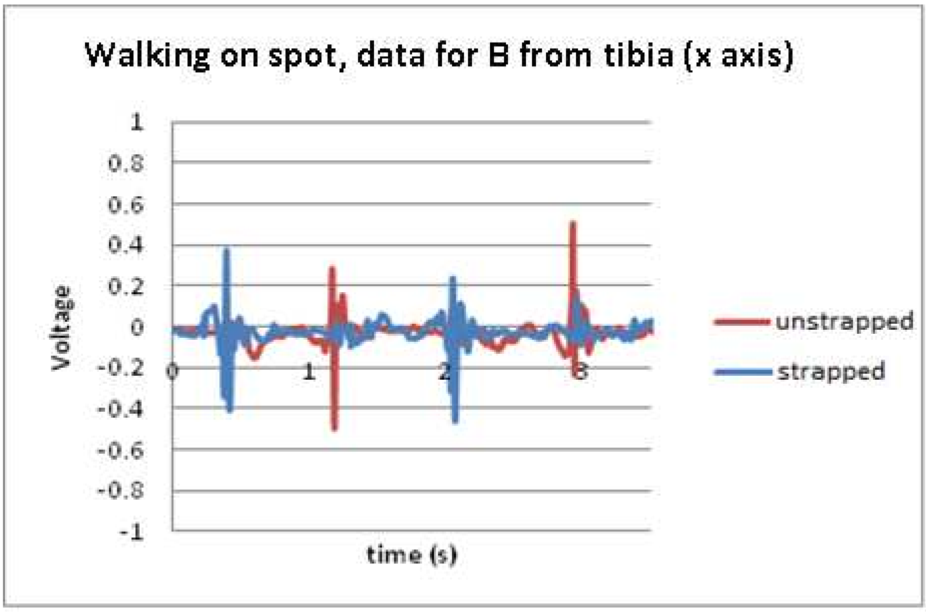
The damped frequency is

From these values, the natural frequency can be worked out.

$$f_n = \frac{f_a}{\sqrt{1 - \zeta^2}}$$

These calculations were solved using formulas in Excel.

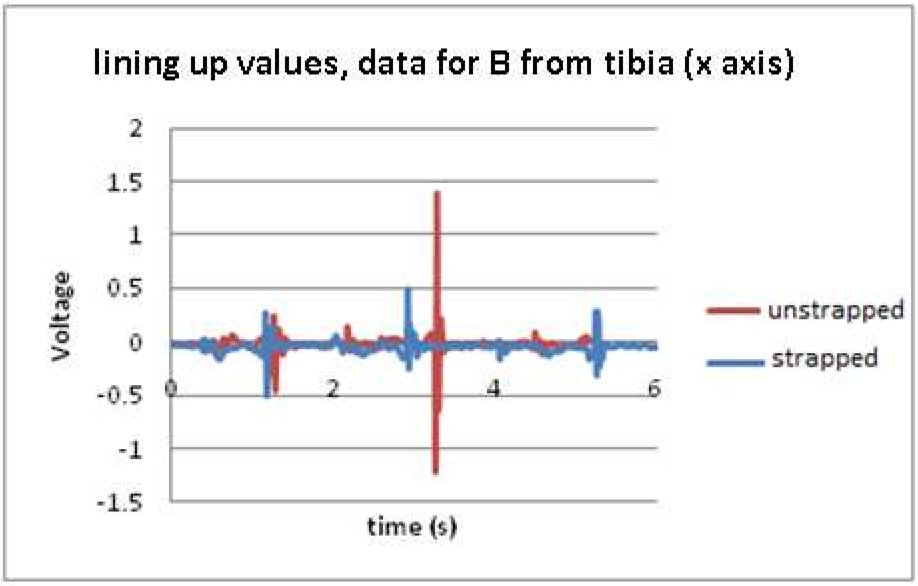
To quantify soft tissue artefact at the tibia, shank and thigh during walking motion, graphs that compared the voltage output at each point were produced. A Graph obtained at this point can be seen in Graph 3.2. From this, the ground force reactions were lined up for each leg, so as to compare movements occuring under loaded and unloaded conditions. This can be seen in Graph 3.3.



Graph 3.2.

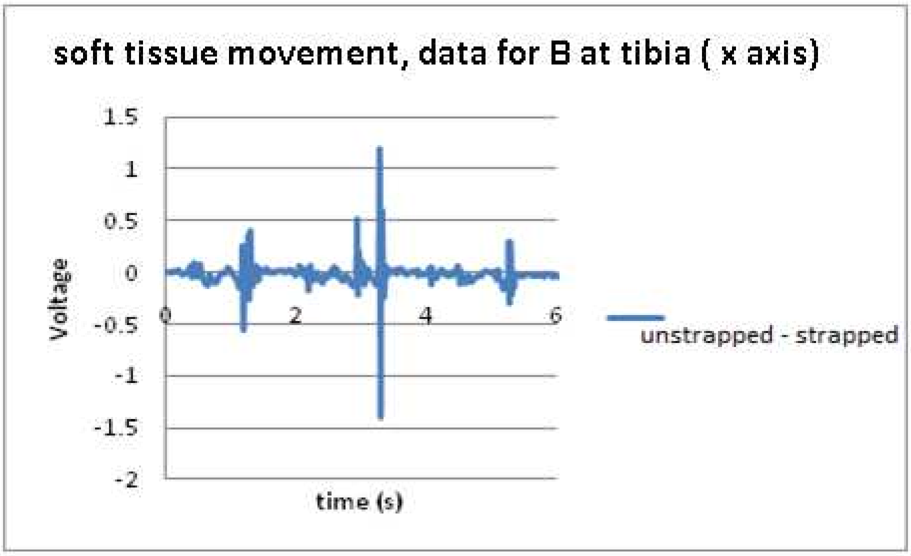
Lining up the data by high values the ground reaction forces gives the results seen in Graph 3.3.

Graph 3.3.



From here, a subtraction of the loaded data was applied to the unloaded, which resulted in a voltage output that represented soft tissue movement. This can be seen in Graph 3.4.

Graph 3.4



Once a voltage reading was aquired for the soft tissue movement, that data was calculated into values of acceleration at each point. The voltage supply was 5V, the sensitivity of each accelerometer was quoted from the data sheet to be 100mV/*g*. The acceleration at each point was calculated by the equation:

$$acceleration = \frac{V_{out}}{0.1} * 9.81$$

Applying a double integration, to work out the velocity and displacement whereby:

$$velocity = \frac{a_1 + a_2}{2} * 0.001 + C$$

$$displacement = \frac{v_1 + v_2}{2} * 0.001 + C(a) + D$$

Where C and D are intergration constants.

This process was repeated to work out the soft tissue movement at the tibia, shank and thigh for each participant.

It should be noted that prior to this analysis, this method of calculating the displacement from accelration was tested with a single accelerometer being moved along the measured distance of a ruler. This revealled that *C(a) + D = 0,* and so the intergration constant was zero.

CHAPTER 4. Results

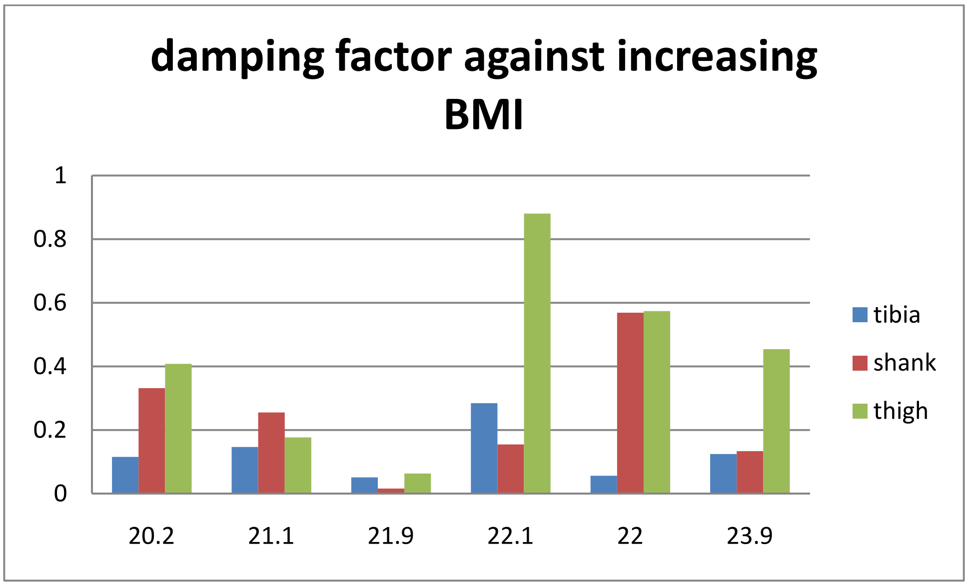
4.1. Nudge test

The values calculated for the damping factor ζ, the damping frequency *fa* and the natural frequency *fn* were acquired at the tibia, shank and thigh of the 6 participants. These values are displayed in Table 4.1.1. Progressing from the tibia to the shank to the thigh, the values obtained for the damping factor increases. Results for the damping and natural frequencies show that progressing from the tibia to the shank to the thigh, values obtained at each point decrease. These patterns for the damping coefficients can be seen in the bar charts below. The body mass index of each participant is charted against the damping coefficients, to determine a relationship between this and the properties of each point.

Table 4.1.1.

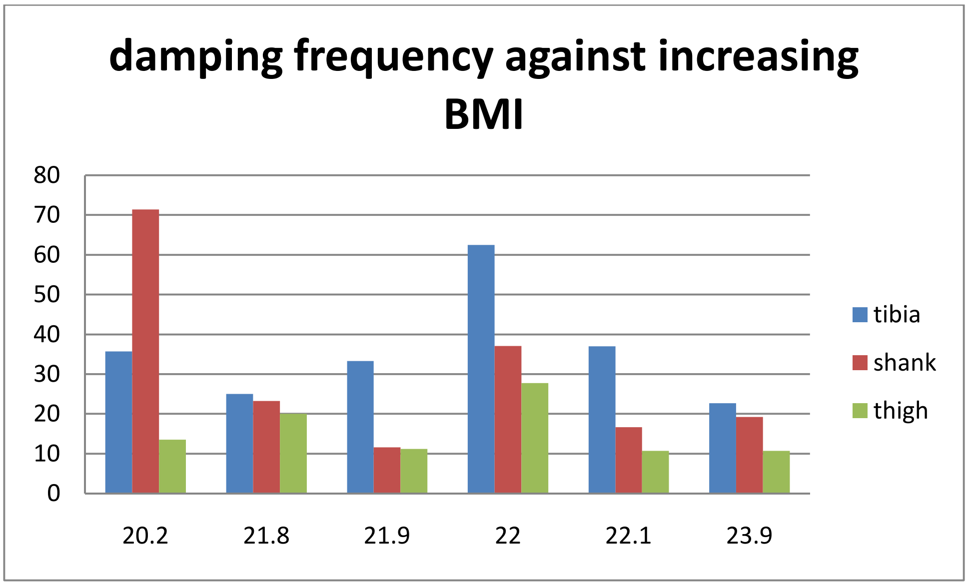
Data obtained at nudge test									
	Properties at the tibia			Properties at the shank			Properties at the thigh		
subject ***(BMI)***	*ζ*	*Fa (Hz)*	*Fn (Hz)*	*ζ*	*Fa (Hz)*	*Fn (Hz)*	*ζ*	*Fa (Hz)*	*Fn (Hz)*
A *(22.1)*	0.147	37.03	37.442	0.2551	16.67	17.23	0.1772	10.75	10.92
B *(23.9)*	0.12502	22.72	22.907	0.134071	19.257	19.405	0.45435	10.752	12.071
C *(22)*	0.057	62.5	62.603	0.5692	37.05704	45.049	0.5741	27.78	32.45
D *(21.8)*	0.05118	25	25.0316	0.0163	23.255	23.258	0.064	20	20.04131
E *(20.2)*	0.116	35.71429	35.957	0.33182	71.42	75.19	0.4082	13.51351	14.803
F *(21.9)*	0.2847	33.33	34.77	0.1551	11.62	11.7704	0.88	11.23	23.671

Table 4.1.1. Shows the damping factor, damping frequency and natural frequency obtained from the nudge test at the tibia, shank and thigh regions.



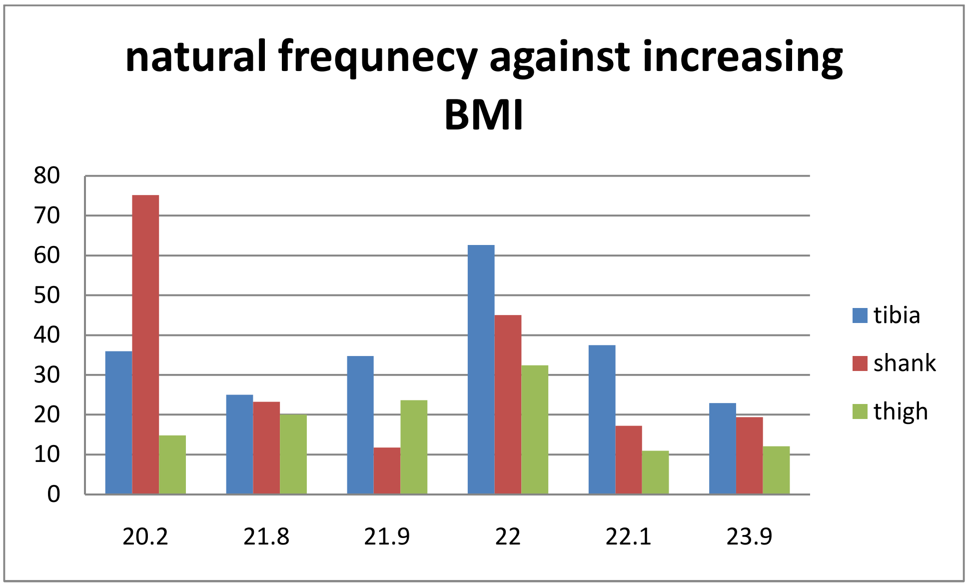
Graph 4.1.2.

In most cases, the damping factor obtained at the thigh for each participant is larger than the shank or tibia.



Graph 4.1.3.

Corresponding to the above the results, the damping frequency, in general, decreases from the tibia to the shank to the thigh. It should also be noted that the participant with the highest BMI exhibits the lowerst damping frequnecy at the thigh.



Graph 4.1.3.

The natural frequency conveys a similar pattern to that of the damping frequency. Graph 4.1.3. shows that in general, the tibia has the highest frequnecy and the thigh has the lowest.

4.2. Comparing values of displacement that were obtained at heel drop

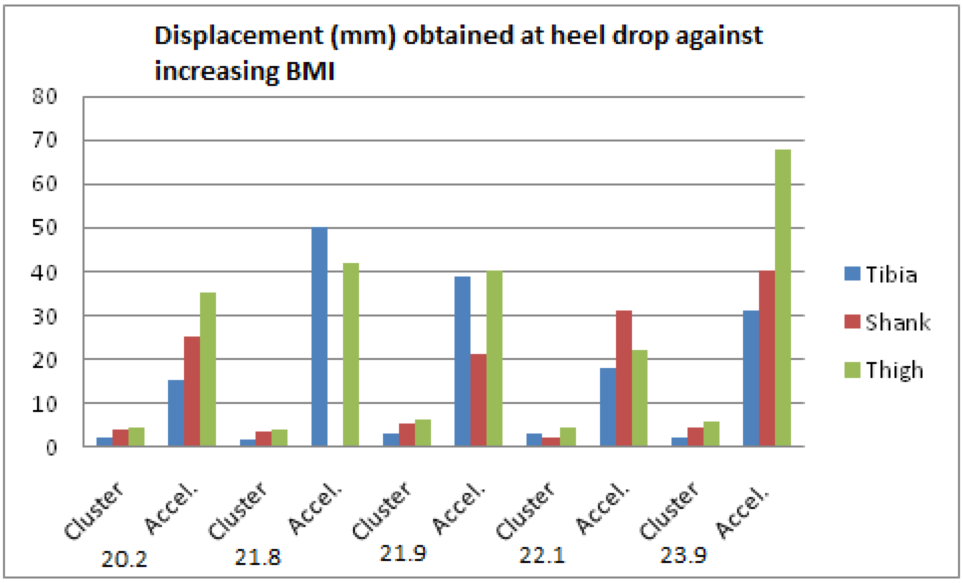
Data obtained at the heel drop for each for the tibia, shank and thigh can be see in the Table 4.2.1.

Table 4.2.1

Comparing soft tissue movement obtained at heel drop from point cluster technique and accelerometers										
Subject (BMI)		*Displacement at the Tibia (mm)*			*Displacement at the Shank (mm)*			*Displacement at the Thigh (mm)*		
		Z	*Y*	*Z*	*Z*	*X*	*Y*	*Z*	*X*	*Y*
		D- P	*M-L*	*A-P*	*D- P*	*A – P*	*M -L*	*D- P*	*A – P*	*M -L*
A (22.1)	*Cluster*	4.25	4/51	2.75	5	3.75	3	7	6	4.75
	Accel.		n/a				n/a			n/a
B (23.9)	*Cluster*	2	5	3	4	3.5	3.5	4.5	3	5.5
	Accel.	15	n/a	11	25	19	n/a	35	38	n/a
C (22)	*Cluster*	1.5	7.5	3	3.4	5	3.5	4	2	4
	Accel.	50	n/a	33			n/a	42	15	n/a
D (21.8)	*Cluster*	3	1.5	5.5	5	6	3.5	6	4.5	4
	Accel.	39	n/a	22	21	25	n/a	40	35	n/a
E (20.2)	*Cluster*	2.75	2.7	4.5	2	3	9.5	4.5	4	4.5
	Accel.	18	n/a	22	31	22	n/a	22	35	n/a
F (21.9)	*Cluster*	2	4.5	3	4.5	2.5	3.5	5.5	5	4
	Accel.	31	n/a	27	40	19	n/a	68	37	n/a

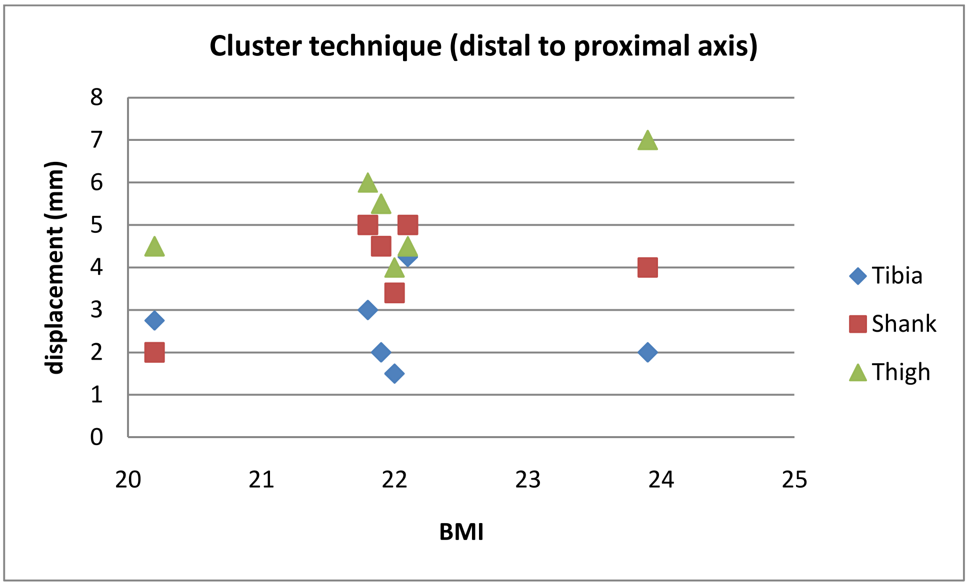
Table 4.2.1, where D-P is the distal to proximal, A-P is the anterposterior and M-L is the mediolatreral axis.

Table 4.2.1. reveals a large discrepancy in magnitude between results obtained by the cluster method and by the use of accelerometers. This difference is highlighted in Graph 4.2.2. Placement of the accelerometers and markers can be seen in Figure 3.1.



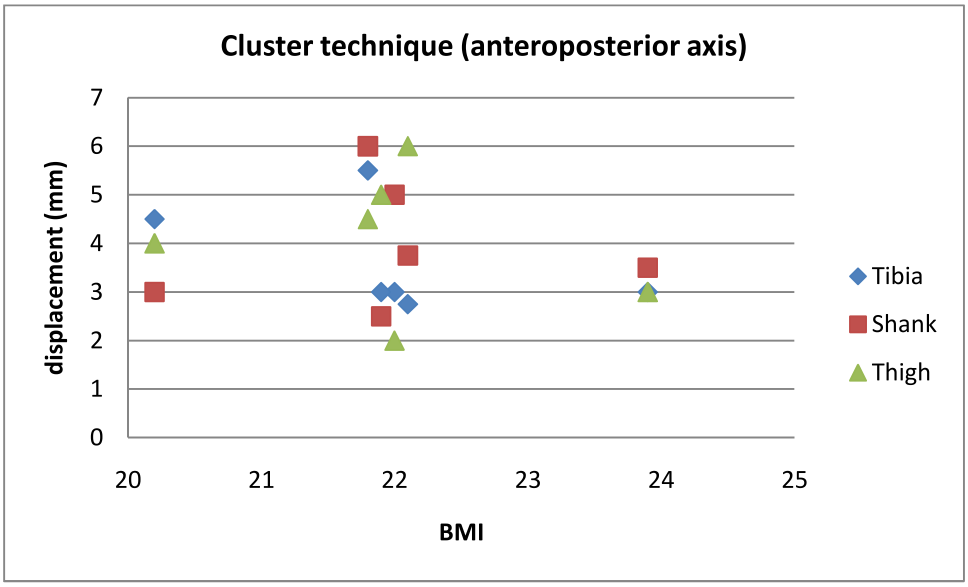
Graph 4.2.2.

Looking closely at the cluster technique, there appears to be a definitive relationship between the low displacements that occur at the tibia and the higher displacements that occur at the shank and thigh. Considering affects that the paticipant's BMI could have on displacement, larger soft tissue displacement occurs in partipant's with a higher BMI. This is especially prevalent at the marker on the thigh.



Graph 4.2.3.

Results in the anteropostrior axis show displacements at each point are more varied, there is less distinction of any pattern that occurs.



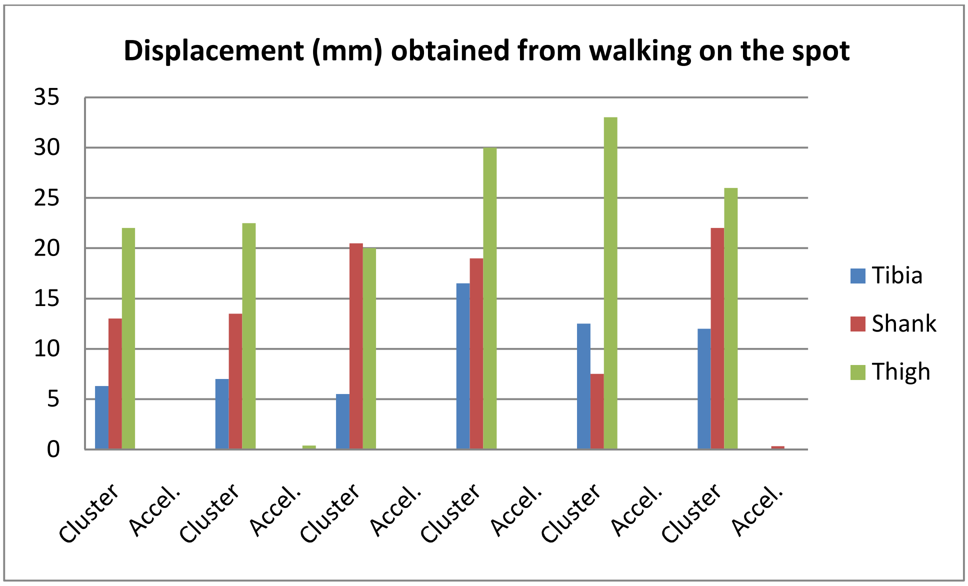
Graph 4.2.4.

4.3. Walking on the spot

The values calculated for soft tissue displacement at the tibia, shank and thigh by means of the cluster technique and through use of accelerometers can be seen in Table 4.3.1. The data shows a large descrepancy between values obtained from each technique. The largest value of displacement using the cluster technique was found to be 30.5mm which occurred at the thigh along the anteroposterior axis. This compares to the far smaller value of 0.069mm that was determined at the thigh across the distal to proximal axis through use of accelerometers and subsequent determining displacement technique. The difference is highlighted the Graph 4.3.2.4. It should be noted that the displacements obtained through use of accelerometers are so small that compared to displacements obtained through use of clusters they cannot be seen.

Soft tissue movement obtained from walking on the spot										
Subject (BMI)		*Displacement at the Tibia (mm)*			*Displacement at the Shank (mm)*			*Displacement at the Thigh (mm)*		
		Z	*Y*	*X*	*Z*	*X*	*Y*	*Z*	*X*	*Y*
		D- P	*M-L*	*A-P*	*D- P*	*A – P*	*M -L*	*D- P*	*A – P*	*M -L*
A (22.1)	***Cluster***	6.3	1.7	5.6	13	11.5	8	22	30.5	21
	Accel.	0.035	n/a	0.006	0.06	0.035	n/a	0.045	0.017	n/a
B (23.9)	***Cluster***	7	3	25.5	13.5	23	4.5	22.5	26	5.5
	Accel.	0.024	n/a	0.04	0.064	0.024	n/a	0.4	0.35	n/a
C (22)	***Cluster***	5.5	2.5	12.5	20.5	15	12.5	20	7.5	12
	Accel.	0.054	n/a	0.021	0.051	0.03	n/a	0.069	0.033	n/a
D (21.8)	***Cluster***	16.5	4.5	23	19	22.5	5	30	8.5	4
	Accel.	0.02	n/a	0.015	0.031	0.06	n/a	0.035	0.02	n/a
E (20.2)	***Cluster***	12.5	4	7.5	7.5	11	7.5	33	13.5	6
	Accel.	0.012	n/a	0.006	0.02	0.015	n/a	0.025	0.017	n/a
F (21.9)	***Cluster***	12	3	21	22	17.5	13	26	18	11
	Accel.	0.018	n/a	0.01	0.31	0.03	n/a	0.035	0.015	n/a

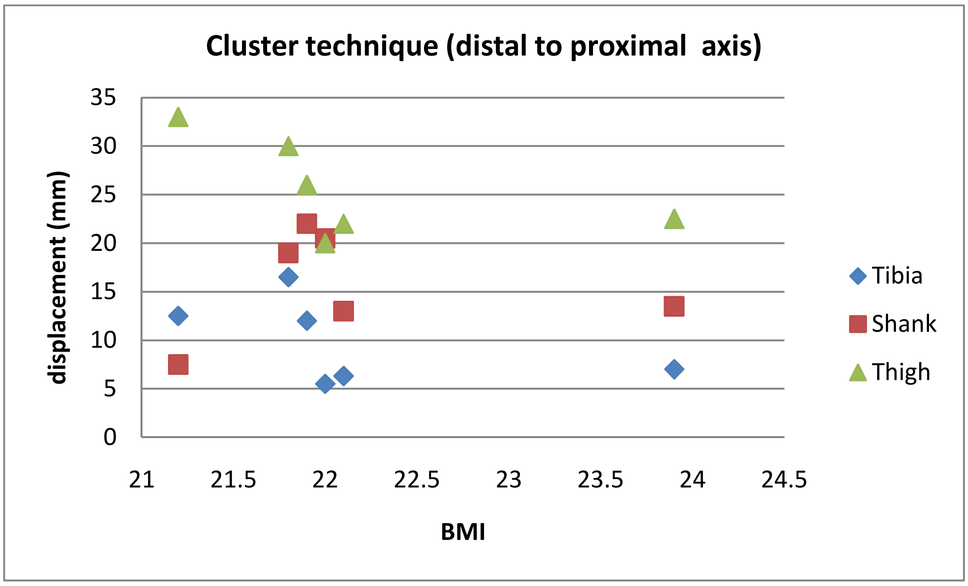
Table 4.3.1. Comparing soft tissue displacement at the tibia, shank and thigh between the VICON cluster technique and the use of accelerometers. D-P is distal proximal, A-P in anterposterior and M-L is medial lateral.



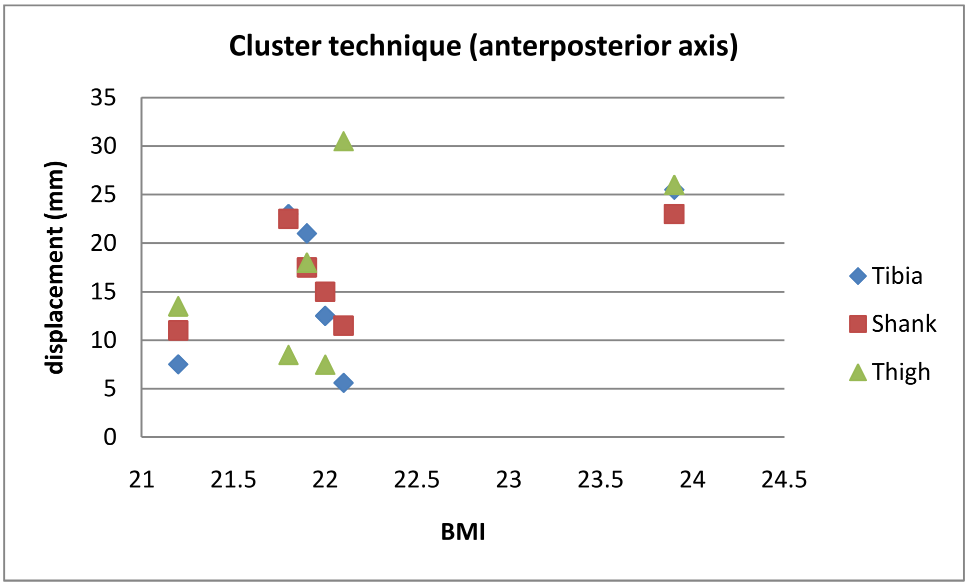
Graph 4.3.2.4

<u>4.3.1 Cluster method</u>

An increase in displacement values can be seen in the progression from the tibia to the shank to the thigh. Graphs 4.3.1.1 and 4.3.1.2 examine the difference in displacements between each region in the anterposterior, dorsoventral and left to right lateral axes. Displacements at the tibia are of a lesser value than those observed at the shank and thigh. Participants with a higher BMI produced larger displacements in the thigh region in particular.



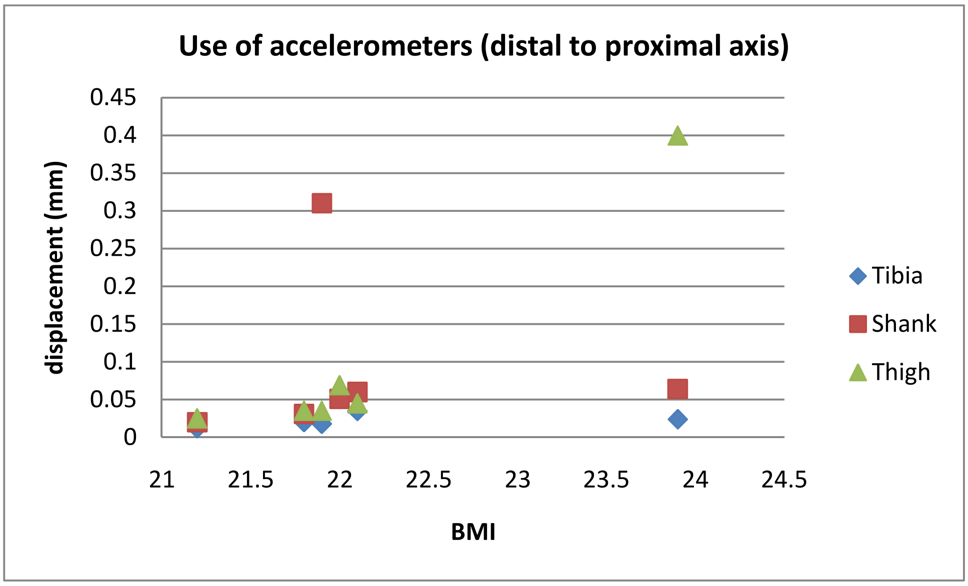
Graph 4.3.1.1.



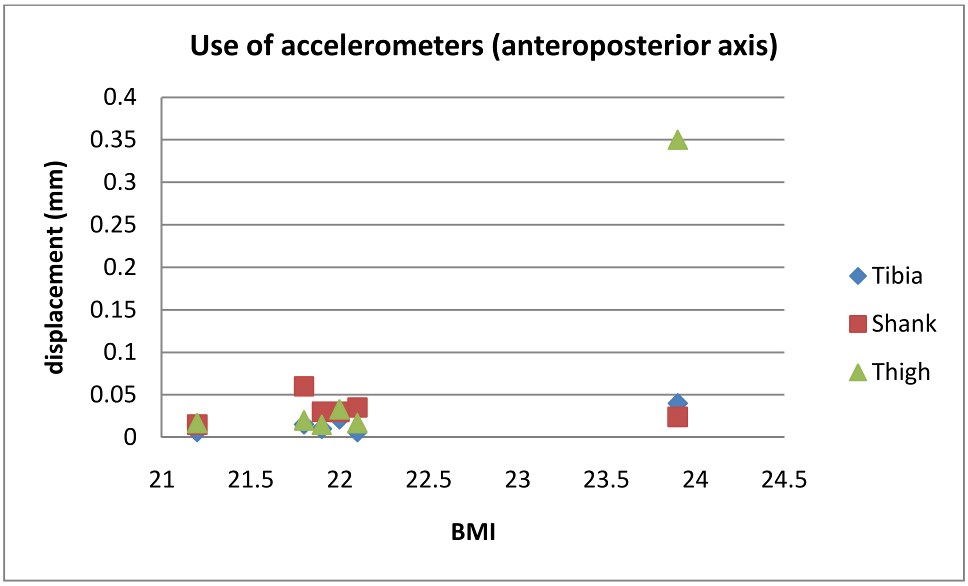
Graph 4.3.1.2.

4.3.2. Use of accelerometers

Less of a pattern emerges from the data obtained from the use of accelerometers in comparison to the cluster technique. Graphs 4.3.2.1. and 4.3.2.2. display indistinguishable results that show, in some cases, displacement at the tibia shank and thigh all having similar values. There is however an increase in the displacement at the thigh for the highest BMI value.



Graph 4.3.2.1.



Graph 4.3.2.2.

4.4. Comparing values obtained from walking on the spot with full gait

Differences in tissue movement at the tibia, shank and thigh from walking on the spot to full gait can be seen in Table 4.4.1.

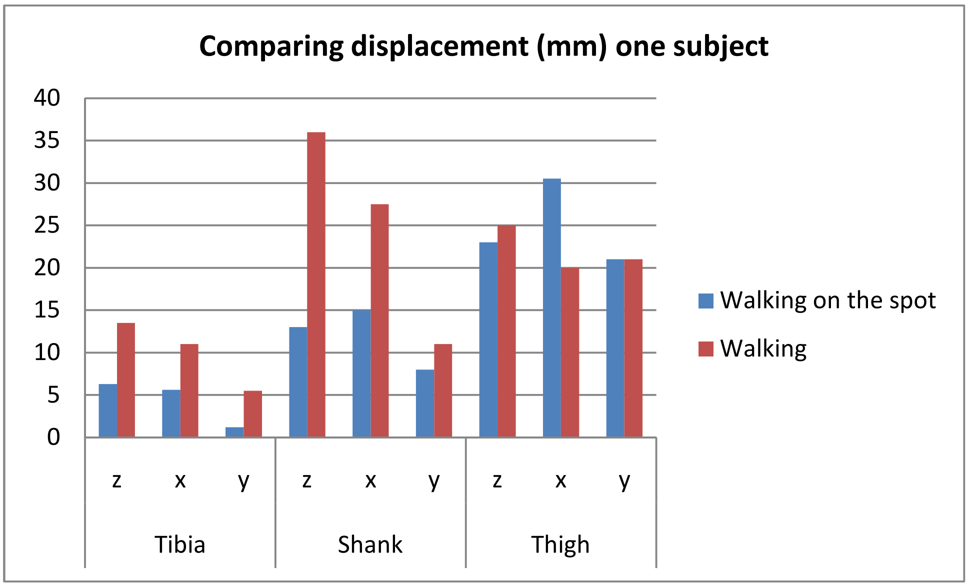
Comparing soft tissue movement obtained from walking on the spot and walking only using cluster technique

Subject (BMI)		*Displacement at the Tibia (mm)*			*Displacement at the Shank (mm)*			*Displacement at the Thigh (mm)*		
		Z	*X*	*Y*	*Z*	*X*	*Y*	*Z*	*X*	*Y*
		D- P	*A – P*	*M -L*	*D- P*	*A – P*	*M -L*	*D- P*	*A – P*	*M -L*
A (22.1)	***W.O.S***	6.3	5.6	1.2	13	15	8	23	30.5	21
	Walking	13.5	11	5.5	36	27.5	11	25	20	21
B (23.9)	***W.O.S***	7	22.5	3	13.5	23	3	23	26	5.5
	Walking	21	40	4	32	30	12.5	20	10	7.5
C (22)	***W.O.S***	5.5	12.5	2.5	20.5	15	12.5	20	7.5	12
	Walking	12.5	25	6	34	20	18	21	7.5	15
D (21.8)	***W.O.S***	14.5	23	4.5	18.5	23.5	5	30	8.5	4
	Walking	19	23	7	30.5	35	8.5	20	17.5	17
E (20.2)	***W.O.S***	12.5	7.5	4	7.5	11	7.5	33	13.5	6
	Walking	17	11	7	30.5	35	8.5	20	17.5	14.5
F (21.9)	***W.O.S***	12	21	3	22	17.5	13	27	18	11

	Walking	21	50	4	17	9	8	68	37	22

Table 4.4.1. Comparing soft tissue displacement at the tibia, shank and thigh considering results obtained during full gait and walking on the spot.

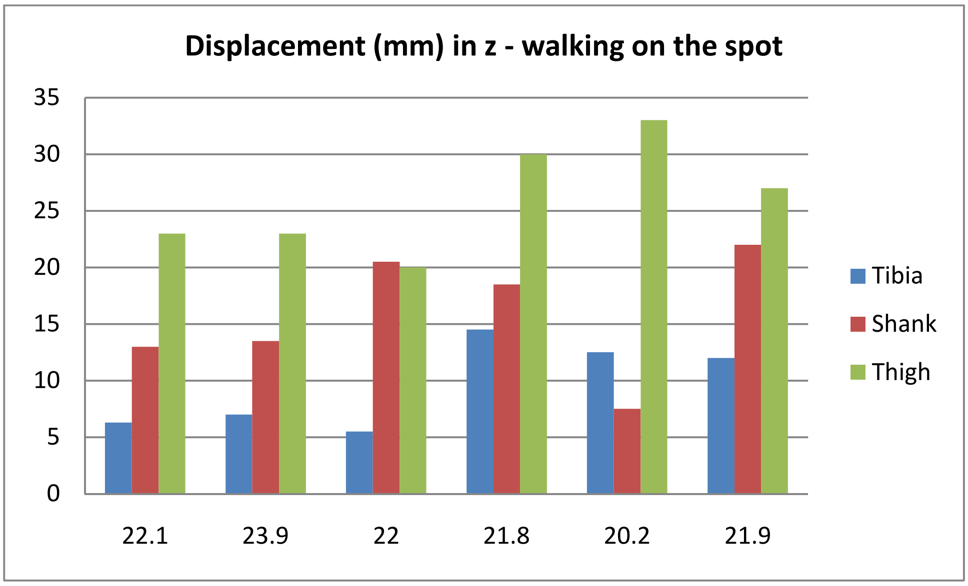
It can be seen that higher displacements were reached during the walking trial than walking on the spot. This difference is highlighted by looking at the displacements of one subject, this is graphed below in Graph 4.4.2



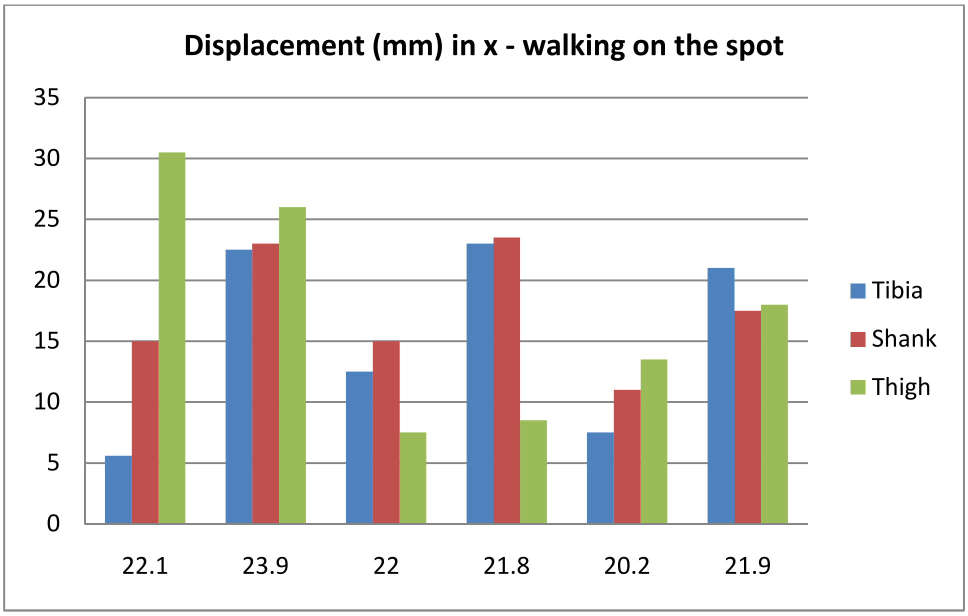
Graph 4.4.2.5

<u>4.4.1. Walking on the spot</u>

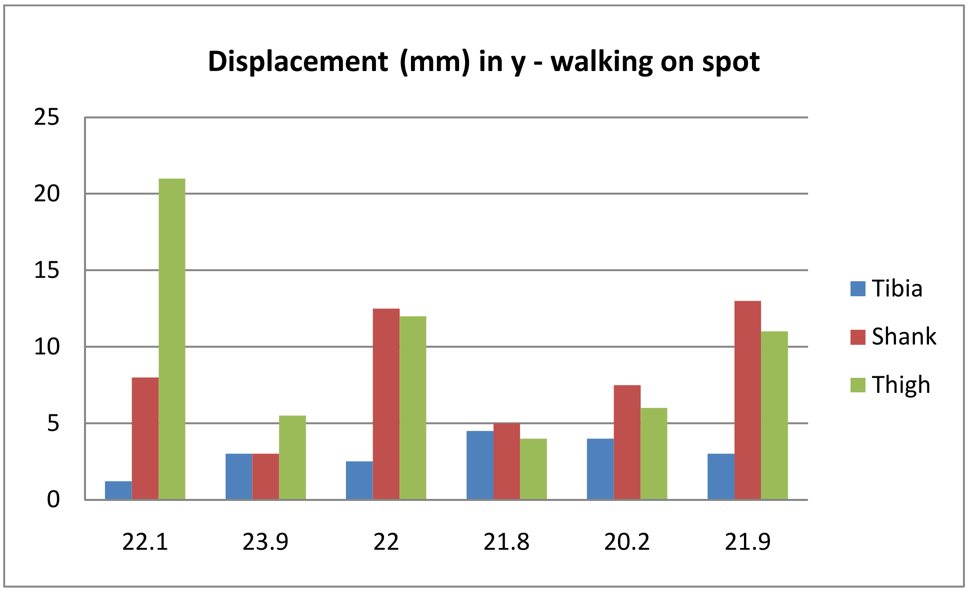
Results show a distinct increase in soft tissue displacement, moving from the tibia to the shank and then to the thigh. Generally, participants with a lower BMI exhibit less soft tissue displacement than those with a higher BMI.



Graph 4.5.1.



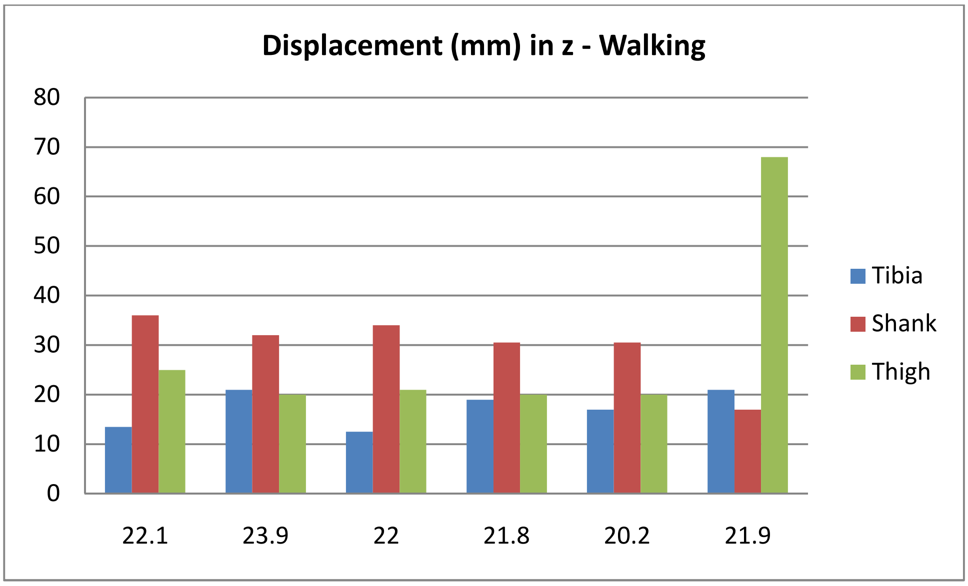
Graph 4.5.2.



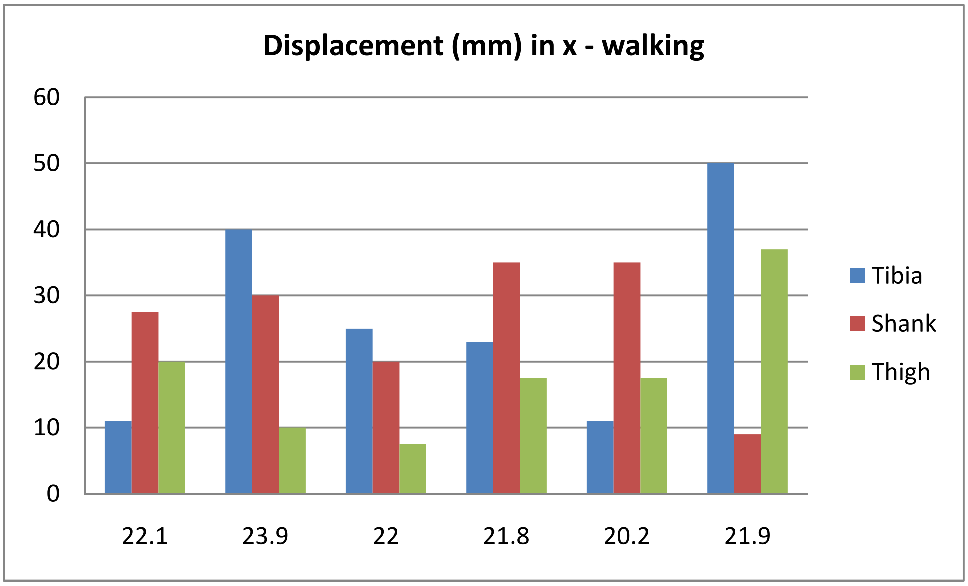
Displacement (mm) in y - walking on spot
25
20
15
10
5
0
22.1
23.9
22
21.8
20.2
21.9
Tibia
Shank
Thigh

Graph 4.5.2

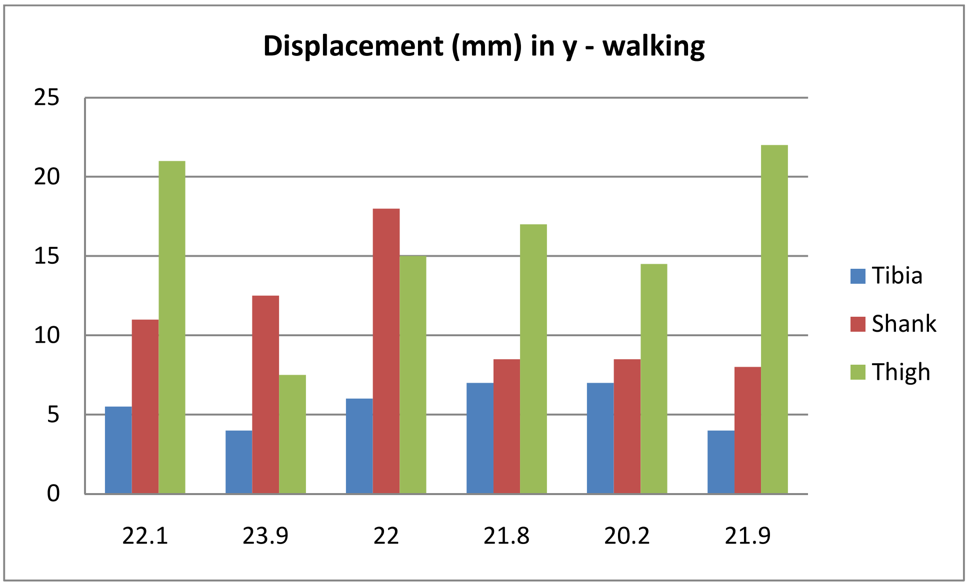
4.4.2. Walking (natural gait)



Graph 4.5.3.



Graph 4.5.4



Graph 4.5.5.

From Table 4.5.1. it can be seen that larger tissue displacements were reached during full gait, in particular at the tibia. This was true for each subject.

CHAPTER 5. Discussion

5.1. Aim of the project

The use of skin based markers in motion analysis can give erroneous results. These errors can be are attributed to soft tissue displacement that occurs over the skeletal frame during movement. Soft tissue exhibits varying viscous and elastic and properties that vary based on the regional aspect of the tissue. This paper explored two techniques that investigated the properties of soft tissue and artefact that occurred in different places during gait. Areas investigated were the tibia, shank and thigh of 6 participants.

5.2 Damping coefficients obtained from the Nudge Test

The skeletal movement and ground force reactions that occur during gait transcends soft tissues and can cause vibration. Properties of the soft tissue dictate damping of the oscillated vibration.

Application of the nudge test provided a means of analysis of the damping coefficients for each region. From the results it can be seen that, in general, the damping factor at the tibia yields the lowest results compared to the shank and thigh. This could be due to the absence of soft tissue that would normally provide damping. It is widely accepted that soft tissue is more prevalent at the thigh and shank than the tibia. This result indicate a relationship between the damping and the amount of soft tissue in the region. Considering the effect of body mass, participants with a higher BMI recorded higher damping factor values, especially in the thigh region. A high BMI is associated with large amounts of body fat.

It was found that the natural and damping frequencies are inversely proportional to the damping factor. Analysis of results reveal that, generally, the tibia exerts higher damping and natural frequencies than the thigh and shank. This is consistent with the nature of large soft tissue areas that tend to "jiggle" at low frequencies. The results correspond to some found by J E Smeathers (1989) who conducted an experiment to determine the damping that occurred across the spine. It was found that the highest values of damping frequencies were found in the thoracic region, an area with the least amount of soft tissue.

Incosistencies in the data, for example exceptionally low values of damping were obtained across all regions for one participant and towering frequency results were obtained at the shank of another, could be caused by varying degrees of nudge applied. Extra care was taken to ensure applied pressure remained the same, however this is tricky to judge and the confines of the experiment did not allow measurement of of pressure.

5.3. Comparing heel drop data obtained by accelerometers and cluster technique.

Soft tissue displacement during heel drop was acquired using the cluster technique and by the use of accelerometers. This allowed for a direct comparison between the two methods and revealed a major discrepancy between both sets of results. Graph 4.2.2 highlights the difference between the two. The highest value of displacement obtained from the cluster technique medial lateral axis. This value is comparatively small to the largest displacement of 64mm that was found in the same region and axis by use of accelerometers.

A study (Lafortune and Lake, 1991) that used bone markers on the tibia, the same spot as this experment, recorded skin displacement values of up to 20mm during knee flexion – extension tasks. In addition to this, Rienschmidt et al. (1997) obtained values at the lateral tibial condyle that presented maximum difference values in the range of 2.6 – 5.8mm from three volunteers. The technique that corresponds more favourably to displacements found via intra-cortical pins is the cluster technique. It is useful to validate measurement techniques by comparing data from areas that are not prone to large soft tissue movements. The tibia, in this case, is the most prominet bony segment and exhibits relatively small amounts of soft tissue artefact. In addition, the lateral aspects of the shank and thigh is a actually a very large region and so validating results at these points is not advised.

The size of the values obtained by use of accelerometers could be explained by the change in distance excised by the entire limb during heel drop motion. This indicates that the accelerometer picks up accelerations recorded on a wider plane than the vibrations of the skin surface. This confirms that readings should be filtered to eliminate walking acceleration, to produce readings of slight movements of soft tissue artefact.

Results obtained by the cluster technique in the distal proximal axis reveal higher values of displacement at the thigh, slightly smaller at the shank, and smaller again at the tibia. Considering the muscles that are engaged during heel drop, it would be expected that most soft tissue displacement would occur in the distal proximal axis.

A relationship between BMI and the soft tissue displacement that occurs for each participant can be established from the results. It appears that as the BMI of each participant increases, so does the soft tissue movement, particularly that found at the thigh. This indicates that confirms that areas with greater soft tissue are prone to further displacement.

The anteroposterior axis contains a smaller range of displacement with a less distinct pattern between movement occurring at the tibia, shank and thigh and any affect BMI has on these.

5.4. Comparing data gathered from cluster technique and accelerometers for walking on the spot

Sach and Lakes (1977) stated that in order to acquire accurate movements of an underlying bone, a skin based accelerometer should be firmly attached to the skin with a preload to compress soft tissue and increase stiffness of the skin. To filter out the walking movements being picked up from the accelerometer, a bandage was wrapped around the segment and values of acceleration found here were subtracted from values gotten by the unloaded accelerometer on the corresponding limb. Examining the results obtained by this method and comparing it to the cluster technique, there are big differences between the range of values of each set. The maximum displacement obtained through use of accelerometers and after processing is 0.069mm, which was reached at the thigh across the distal to proximal plane. Whereas the largest displacement found using the cluster technique was 30.5mm, which occurred at the thigh region across the anteroposterio plane. Comparing these data to that acquired from a technique based on roentgen photogrammetry (Sati et al., 1996) gauges similar results. Markers attached to a region with assumed small instances of soft tissue artefact, the lateral condyles, recorded displacement of 3.47 and 2.93mm medially and laterally for one subject. At the other end of the scale and under the same experimental conditions, another subject accumulated errors of up to 8.9 and 9.13mm, respectively. This indicates the wide range of soft tissue artefact occurring at the same places between different participants. Using these results as a general standard for verification of the two techniques being investigated in this study, the cluster technique is deemed more favourable. Whilst varying degrees of error can be seen between each subject, and this is highlighted in Graphs 4.3.2.1. and 4.3.2.2, all values are contained within 0 - 35mm distally and 0 – 31mm anterposteriorly. Data obtained by the use of accelerometers show that displacement values occur in the range of 0 – 0.045mm and 0.037mm, respectively. Although it would be desirable for the error in skin marker readings to be so low, previous results from numerous studies (Lafortune and Lake, 1997; Sati et al., 1996) found that soft tissue generated greater values of displacement than this. Such small values could indicate that perhaps the control, ie the loaded accelerometer data was too high. This could be down to various reasons. The first, is that readings of acceleration in the limb could be incorrectly exaggerated by a reaction from the ground force upon heel strike. J E Smeathers (1989) found that accelerations obtained at the ankle during heel strike reached $35m/s^2$. Although participants wore soft soled running shoes in an effort to damp the vibration caused at heel strike (McGuire, 2009.), these forces cannot be eliminated. Timescale limitations of the experiment inhibit the ground force reaction values being explored.

A second offering, as to why the use of accelerometers have given uncharacteristic soft tissue displacement, could be due to the bandage technique being an inadequate means of stifling tissue movement. A technique that addressed soft tissue movement by applying an elasticated bandage to the skin (Cordero, Mateu-Arce at al. 2008) was used in the current study, however a inelastic cotton band was used in this case. A diagram from the paper indicates the dimension of bandage used. An attempt to hold the accelerometer tight and not to hinder the movement of the rest of the limb proved troublesome, as the participants expressed discomfort at relatively loose levels of fastening. Soft tissue movement transcends limb segment (Kepple et al., 1994) and the application of a bandage could mean that although tissue directly under the bandage may not relative to it, the entire bandaged area could still move relative to the bone. This could be especially true in areas with large amounts of soft tissue.

A further factor that could have affected the results obtained from the accelerometers is that readings of electrode potential generated by muscles, noise and temperature were taken from each participant at the start of the experiment and subtracted from values that were obtained after walking tasks had been performed.

It should also be noted that the integration technique used in the process of data to gauge values of displacement was only basic. Mathematical rules coincide that integration should contain limits. The limits in this experiment would be the contributions such as temperature, muscle potential during gait and ideally the precise acceleration of the bone. Bone mounted accelerometers would give accurate results, however application on these would be painful and it cannot be justified from an ethical point of view.

Furthermore, it was observed that the stiffness of the wire was generating slight movements at the skin surface that would be deemed uncommon during gait. The presence of wires could have given unrealistic gait results, each participant had to pay particular attention making sure they did not trip or pull out any of the wires.

Looking at patterns that emerge from the cluster technique, in general, the tibia produced the smallest instances of soft tissue artefact during walking on the spot. Thigh readings yielded the highest values of displacement. This pattern can be seen via roentgen photogrammetry (Sati et al., 1994) and is consistent with both the damping coefficients and the results of displacement seen at heel drop.

In the anteroposterior axis, 22mm tibial displacement was found. Which corresponds very closely to those gathered from bone pins (Lafortune and Lake, 1997) wherein by the maximum displacement of 22mm at the tibia during knee flexion/extention was obtained. Maximum test was found at the thigh, along the anteroposteroplane and was 31mm with 32mm found in the distal proximal plane. These correspond nicely to displacements of 33.19 and 38.3mm found at the thigh (Sati et al., 1994) using x-ray fluoroscopy.

Behaviour of the soft tissue artefact found at regions by use of accelerometers are nonconstrued, which reitterates the conlusion that undefined ground forces could be giving adverse results.

Compare soft tissue displacements from walking on the spot with walking

The length of wires attached to the accelerometers presented limitations when attempting to carryout analysis in gait. A compromise was made where the current study looked at the displacements obtained at walking on the spot. To validate this decision, displacements gathered during walking on the spot and a further gait analysis using the cluster technique was analysed. Comparing the results in Graph 4.4.2 it can be seen that there are distinct differences between the displacements reached during normal gait and walking on the spot for each participant. It was observed that full swing is not achieved on the spot, and widely accepted that maximal soft tissue displacements is generated at extremities of motion (Cappello, 1997) Reviewing the Graphs 4.5.1 – 4.5.5 it can be seen that although they are at different scales, similar patterns throughout displacement at the tibia that increases when at the shank and furthermore at the thigh are littered throughout both sets of data. This pattern has been seen throughout this current study, particularly in the distal proximal axis which contains the highest levels of artefact which is the bases for conclusions that were drawn.

CHAPTER 6. Conclusion

It was found that the cluster technique produced more favourable results than data gathered through use of accelerometers when quantifying soft tissue displacement by comparing values gathered with those found in the literature. Results of each experiment were compared at the tibial tubercle, where it is understood that the occurrence of soft tissue displacement is small. It was concluded that unrealistic displacement values, obtained through the use of accelerometers, were due to the amount of uncertain variables such as inconsistencies in the strapping method, irregular values obtained from the electrode potential generated by muscles and crude “lining up” of walking data and the subsequent simplitic double integration applied. The aid of the bandage to minimise the movement of soft tissue during motion was also found to be of limited value in reducing the accelerations.

The cluster technique gave results that were close to those found in the literature. Addressing the problems of skin marker movement relative to the bone, using a method developed in VICON means that errors are contained as no external measuring techniques are being implemented.

A relationship appears to exist between the BMI of a participant and the displacement obtained at areas of large soft tissue, this should be explored in future studies.

As expected, larger soft tissue movement was found at the thigh region than the shank and tibia.

6.1. Suggestions for further study

For future study I would like to investigate further the relationship between BMI and soft tissue artefact. Generating something as simple as a standard table of values that could indicate probable errors occurring at skin markers at selected regions of the body that would be determined by the BMI could greatly help clinicians.

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