USE OF ULTRASOUND WITH MOTION CAPTURE TO MEASURE BONE DISPLACEMENT DURING MOVEMENT MADE FOR FUNCTIONAL HIP JOINT CENTER DETERMINATION

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ABSTRACT

Calculation of hip joint center (HJC) through functional methods with markers placed on skin around thigh and pelvis is a non-invasive method for estimating the center of rotation of a ball and socket joint by recording movements of femur relative to acetabulum. But the HJC through this process suffers from a well-documented source of error known as soft tissue artifacts (STA) which is the major source of error in determining functional HJC. Previous experiments associated with STA determination and compensation for HJC estimation have been invasive such as bone pins and hence are not viable for human based studies. After a thorough study, there appeared to be a need for a non-invasive ad-hoc procedure to quantify this error source. We have conducted a set of experiments to see change in thickness of soft tissues from skin surface on thigh up to bone using ultrasound as an ad-hoc to a motion capture system. In this study our hypothesis was that during the movement of thigh the bone moves linearly with respect to the marker on skin in the direction of probe and depth of bone from skin surface changes linearly in the direction of movement. Motion type "Flexion" with bent knee showed a maximum bone displacement of 1.5cm from neutral position with respect to skin with a maximum relative displacement of a virtual skin marker by 27cm and a correlation 0.865 in synchronized frames.

INTRODUCTION

Functional HJC is a well-documented method to find center of rotation of hip with help of external markers [1][2][6].The non-invasive and easy implementation of the experimental procedures, along with results close to the true hip center in human studies [7][8][12], have made this method attractive over others for gait analysis as well as for determination of a reference point in navigation based surgeries[6]. Studies using Virtual simulations [2][6] and Mechanical linkage[2][4] give accurate results within 1mm of error showing accuracy of algorithms. Although, when similar algorithms are used in vivo on humans, the error rate increases considerably up to 20 mm as reported by a recent study on humans by Sangeux et al [8]. It is indicated by Heller et al[11] that these errors in humans are coming from soft tissue component which is missing in mechanical linkage or simulation data. This source of error is reported to have frequency content similar to bone movement and hence cannot be removed using signal processing or filtering [14].

Statistical methods such as Procrustes Analysis have been used to get an estimate of STA non-invasively [11][13]. Although in our knowledge there were no studies found to quantify the reason behind soft tissue artifact through non-invasive procedure using ultrasound. Hence in order to identify how the underlying bone is moving with respect to the skin where the markers are attached which might affect the calculation of HJC using the reconstructed poses from the markers, this experiment was conducted and it was hypothesized that ultrasound could be a possible ad-hoc addition to functional HJC calculations which can give real time bone movement information with respect to skin while the standard movements [2] are made. Ultrasound is low-cost and safe imaging modality which has been used recently to validate functional HJC providing gold standard data [8][15]. Hence it was presumed that femur bone data and its depth variation might be visible in real time motion through ultrasound.

METHODS AND MATERIALS

Four human subjects participated in the study. Setup consisted of ultrasound imaging machine (Picus, Esaote Europe) and linear probe (L10-5, 5 MHz operating frequency, width 4 cm). The motion capture system consisted of 6 VICON MX40 cameras at the frame rate of 120 Hz. 9 retro reflective markers were used, 3 each on thigh and back and 3 on probe with an extension to track the position of probe movement.

The participant held the probe and stood upright for the neutral pose as seen in Figure 1. For motion type, Flexion Bend (with bent knee) probe was placed vertically (Probe’s longer edge parallel to the bone) at front and side on the thigh. The movement was started with a quick movement perpendicular to the bone to synchronize the motion data with ultrasound along with time stamps. After the jerky movement the participant flexed the leg from hip with bent knee, made it reach the maximum of their caliber and then returned it back to the neutral pose. The ultrasound recording was started with the jerky movement up to 6 seconds as the limit for ultrasound machine was to capture at 30 Hz for total 180 frames. The VICON motion capture was started before ultrasound measurement while participant stood still and was stopped only after ultrasound recording was stopped.

Figure 1 : Setup with participant handling the ultrasound probe. Ultrasound machine was covered with cloth to avoid reflections and VICON camera

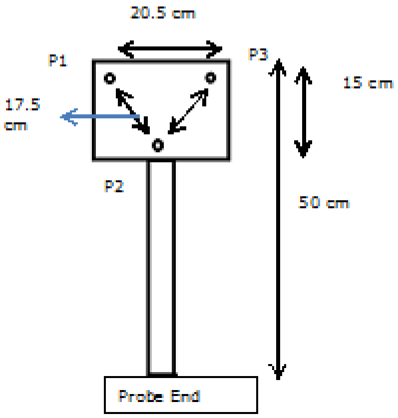

Figure 2 : Ultrasound probe and marker attachment.

CALCULATIONS OF TISSUE THICKNESS WITH FLEXION

The motion data was analyzed in terms of relative displacement of a virtual marker placed on skin. This marker position was calculated with help of 3 markers placed on the ultrasound probe. For each frame, P1,P2 and P3 were three markers on probe in Figure 2 and their *x,y,z* coordinates were obtained. The direction perpendicular to P1P3(vector) towards P2 was calculated. This vector ,v was used to translate P2 in space by 350 mm(Distance of marker P2 from thigh surface), to reach the surface of skin on thigh. Its relative displacement was then calculated wrt the position in neutral pose. Frame 1 was considered to be neutral pose and hencethe displacement is

$$\sqrt{(((x_i-x_1)^2+(y_i-y_1)^2+(z_i-z_1)^2))}$$

i = 1 to N, where N is total number of frames captured and x,y,z are coordinates of calculated marker position on thigh.

For ultrasound data, the surface of the bone was visible as a bright intensity band against noisy speckled background. The edge tracking software “EdgeTrak”[5], was used to get a set of open contour points which provide the position of bone with respect to the skin surface. All the ultrasound data consisted of 180 frames and 100 contour points were generated for each frame using a scaling factor which converted pixels to mm. From this contour data, variation in depth of edge of bone was calculated using mean of y coordinates for each frame. Relative displacement of this depth with respect to the first frame was reported. The first frame was considered as neutral position of depth of bone in standing pose. The initial jerk given to ultrasound probe generated a spike which was considered for synchronization with VICON data.

Synchronization between ultrasound and VICON

The synchronization was made through analysis of graphs while the starting point of movement was considered with an increasing slope in VICON data and after spike in ultrasound data. Numbers of frames were converted to time domain using the conversion of 30Hz for ultrasound and 120Hz for VICON data. Every 4 samples of VICON data contained 1 ultrasound sample. Rough approximation was made using stamping in the graph.

RESULTS

Maximum displacement of bone with respect to neutral position in terms of depth from skin on thigh and maximum relative displacement of virtual marker placed on skin where probe was placed are reported in Table 1. For synchronized data, it was observed that the variation in soft tissue depth and movement were related. Figure 3 shows that while the depth of bone decreases (relative displacement increases in direction of probe) as the flexion increases up to a maximum and then increases (relative displacement decreases in direction of probe) in the reverse motion. The correlation values obtained between marker displacement and bone displacement are in Table 2. It was observed that an initial rise in displacement occurred while the probe was placed on side (lateral side) too. Maximum displacement in this direction was observed to be half of that in front for two participants (1 and 4). Average displacement of bone was much lesser when probe was placed on side than in front. In Literature, one of the methods to quantify the soft tissue artifact was reported as displacement of marker attached on skin with respect to marker attached on a pin inserted into the cortical bone [14]. This reached up to 10 mm in the study reported by Leardini et al [14]. In our study we have quantified a similar metrics with non-invasive ultrasound and the maximum displacement was around 15 mm.

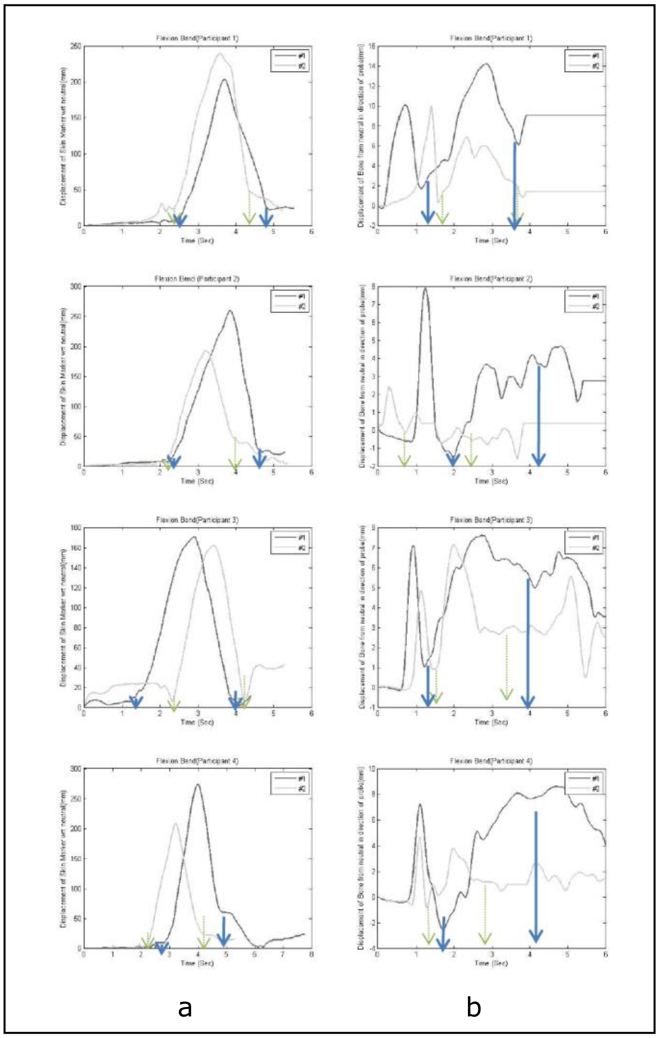

Figure 3 : a) Relative displacement of skin markers in 3D space from VICON, b) Relative Displacement of bone with respect to skin through Ultrasound. Legend: #1(Black): Data with Ultrasound probe at front on thigh, #2(Gray): Data with ultrasound probe at side on thigh(lateral).

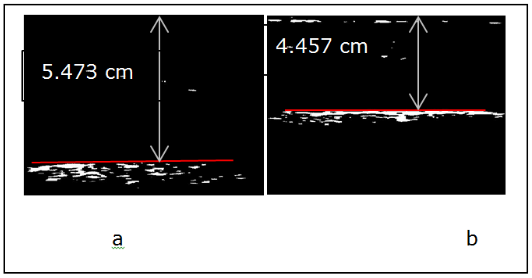

Figure 4 : Ultrasound image for participant 4 with probe at front thigh a)neutral pose b)flexed pose. Thickness of soft tissue is the distance between skin and surface of bone(image is processed to get boundary of bone)

Table 1: Displacement of skin marker and bone with motion type "Flexion (Knee Bent)"

Participant	Probe location (On thigh)	Maximum Marker Movement in space (cm)	Maximum Bone Movement wrt skin (cm	Average Bone Movement wrt skin (cm)
1.	Front	27.337	1.486	0.94
	Side	20.909	0.705	0.32
2.	Front	25.896	0.476	0.216
	Side	19.256	0.038	-0.020
3.	Front	17.091	0.779	0.608
	Side	16.229	0.731	0.370
4.	Front	27.337	0.866	0.486
	Side	20.909	0.414	0.161

Table 2 : Correlation of synchronized data from Ultrasound and VICON (for bone and marker on skin)

Participant	Probe location (On thigh)	Correlation ($P < 0.001$)
1	Front	0.865
	Side	0.897
2	Front	0.525
	Side	-0.242
3	Front	0.609
	Side	0.737
4	Front	0.699
	Side	0.537

LIMITATIONS

Ultrasound data was noisy and some of the frames were missing due to misplacement of probe during the motion. These frames were manually identified and the value was treated as an outlier with mean value treatment. Ultrasound data for participant 2 were very noisy with frames missing the bone edge for more than 100 frames out of 180 with probe facing side. A sharp rise in participant 3 with probe facing side is observed against the trend. This is assumed to be contributed to the displacement of probe during motion.

The probe attachment was heavy making it difficult for participant to hold it rigidly during the motion. Also, synchronization is done based on manual observation and analysis of graph based data. In future these limitations are expected to over-come by attaching the probe through a foam based attachment rigidly onto the thigh and improvising automatic synchronization based on time stamps or an external trigger.

DISCUSSION

In Leardini et el[14], it was mentioned that skin markers are not appropriate for estimation of underlying bone. Our experimental study has proved that during one of the motion type, Flexion, the underlying bone position is not constant to the skin at all times. Rather, the bone displaces linearly with the motion from its neutral position in the direction of movement upto 15 mm with our 4 human subjects. This seems in line with cadaver studies [9] performed with transcutaneous bone pins or intracortical pins [10][14] which have shown that there is displacement up to 10mm between the markers attached on skin and the one directly on bone. Moreover the movement of bone in the direction perpendicular to direction of motion was almost half. This data suggests ultrasound could be a useful tool to assess soft tissue displacement and since linear movement is observed, algorithms could be proposed to translate the marker at each time instant to compensate for the bone movement to get a better estimation of underlying bone and hence HJC. Future study will assess other motion types like Abduction and circumduction which are used to locate HJC and possibility of algorithms to compensate STA based on ultrasound data will be explored.

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