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COMPARING SQUAT KINEMATICS AND KINETICS USING SUBJECT-SPECIFIC PELVIS LANDMARKS AND HIP JOINT CENTRES FROM CT DATA

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SUMMARY

We implemented a subject-specific CT approach to better estimate the hip joint locations in motion analysis. Squat kinematics and kinetics were examined using subject-specific hip joint centres and marker positions, located from CT data. This demonstrated slightly higher peak joint angles and moments, compared to the conventional Davis model.

INTRODUCTION

A common approach in motion analysis consists of defining the location of the hip joint centre (HJC) based on Davis' predictive regression models [1], using pelvis landmarks and a function of leg length. The accuracy of this measure has been often disputed [2], since it relies on correct marker placements and does not consider subject-specificity or hip pathologies. Alternatively, the HJC has also been estimated using imaging data and functionally locating the centre of the femoral head [3,4]. We hypothesized that by implementing a subject-specific CT approach, the conventional predictive equations can be modified to better estimate the HJCs and thus provide better kinematics and kinetics. The purpose of our study was to understand the discrepancies of these predictive methods, by comparing resultant squat kinematics and kinetics from two models: (1) Davis, based on the conventional predictive regression model; (2) CT markers, based on locations of the subject-specific surface markers used in motion analysis.

METHODS

Ten (7 males, 3 females) healthy control participants were recruited for pelvic CT imaging and motion analysis (age = 34.7 ± 5.6 years; BMI = 24.6 ± 2.2 kg/m^2). The participants had no previous history of lower-limb surgeries, pathologies, osteoarthritis, or severe injuries. Prior to CT imaging, contrast markers were placed on the left and right anterior superior iliac spines as well as on the left and right posterior superior iliac spines, denoting the anatomical marker locations. The participants were scanned in a supine position with the contrast markers visible on the CT images. After imaging, the contrast markers were removed and replaced with retro-reflective markers. Additional retro-reflective markers were attached to the rest of the body according to a modified Helen-Hayes marker set [5].

Each participant was instructed to perform five maximal dynamic squats. Three-dimensional kinematics were collected using ten Vicon MX-13 cameras (Vicon, Los Angeles, CA, USA) and ground reaction forces were recorded using two force platforms (Model FP4060-08, Bertec Corporation, Columbus, OH, USA). Participants' CT data were blinded, randomized, then read using ITK-SNAP 2.4 (PICSL, Pennsylvania, PA, USA)[6]. Each CT set was evaluated for locations of the contrast markers as well as the left and right HJCs in a three-dimensional x-y-z coordinate system. Three evaluators performed the measurements, each performing three readings. Intraclass correlation coefficient (ICC) values were determined using SPSS Statistics v.20 (IBM Corporation, Armonk, NY, USA) to examine the intra- and inter-observer reliability to evaluate the CT data. Using the locations of the CT markers, the conventional equations from Davis [1].

$$HJC_X = [-x_{dis} - r_{marker}]cos\beta + Ccos\theta sin\beta \qquad \text{Eq. 1}$$
$$HJC_Y = \left[Csin\theta - \frac{d_{ASIS}}{2}\right] \qquad \text{Eq. 2}$$
$$HJC_Z = [-x_{dis} - r_{marker}]sin\beta - Ccos\theta sin\beta \qquad \text{Eq. 3}$$

They were modified to incorporate subject-specific marker locations to determine the left and right HJCs. In our approach, the function *C* was determined from measuring the planar distance between the anterior superior iliac spine and the location of the ipsilateral HJC, located from CT data, as opposed to conventional methods using linear regression equations [1]. The horizontal distance between the anterior superior iliac spine and HJC in the sagittal plane (x_{dis}), and the alignment (θ) and inclination (β) angles were also subject-specific values used to determine the HJCs. The distance between the left and right anterior superior iliac spines (d_{ASIS}) and marker radius (r_{marker}) were included. The squat motion was processed and compared using the two models: (1) Davis, using regression equations; and (2) CT markers, using marker locations. Values for each location were averaged from all evaluators' readings and processed. Root mean square error (RMSE) was determined to examine differences in predicted hip joint angles and moments.

RESULTS AND DISCUSSION

The evaluations of the CT data were reliable among each observer (ICC_{Obs1} = 0.997; ICC_{Obs2} = 0.972; ICC_{Obs3} = 0.963) and between observers (ICC_{Obs1-2} = 0.953; ICC_{Obs1-3} = 0.922; ICC_{Obs2-3} = 0.910). Examining the squat cycle for the two models (Figure 1), both models demonstrated relatively similar hip angles in flexion, abduction, and internal rotation

(RMSE = 1.9, 0.2, 5.5, respectively). The models also demonstrated similar joint moments in the sagittal plane (RMSE = 23.8). The CT markers model was consistently higher and overestimated the joint moment at maximal squat, demonstrating a higher peak hip flexion (111.6°), thus, a higher peak moment during maximal squat (-847.8 N·mm/kg). Differences in joint moment were noticed during quiet stance in the frontal plane, at the beginning and end of the cycle. There were also noticeable differences in the transverse plane, in terms of internal rotation at maximal squat and joint moments during the ascending and descending phases. Table 1 summarizes the means from the resultant peak hip angles and moments, as well as the percentage difference and RMSE between both models in the sagittal, frontal, and transverse planes.

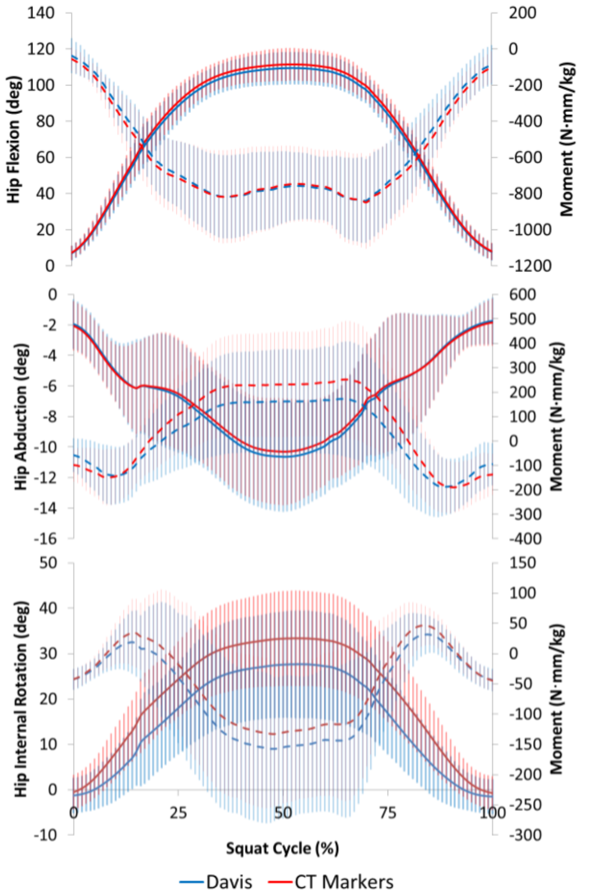

Figure 1: Hip joint angles (solid lines) and moments (dashed lines) in the sagittal (top), frontal (middle), and transverse (bottom) planes during maximal squat – using the Davis (blue) and CT markers (red) models.

Caution should be taken to avoid misleading joint angles and moments. There is a slight uncertainty with imaging participants in the supine position, since the examined joints were not load-bearing and pelvic inclination could have been modified. Incorrect marker placements can misinterpret the HJCs and propagate errors. To yield a more subject-specific representation, locations of the actual bone landmarks should be considered and incorporated to evaluate kinematics and kinetics.

CONCLUSIONS

The ICCs confirmed that the CT measurements were precisely and accurately located (ICC > 0.8). It was evident that the marker positions from the CT data could have overestimated of the joint angles and moments. This was somewhat negligible in the sagittal plane, where the overestimations were relatively small compared to the large amplitudes of hip motion.

Radiographic imaging can be very beneficial towards determining the geometric HJCs [3,4]. The distance between each marker and ipsilateral bone landmark, determined from subject-specific CT data, could be included as a set of virtual markers during dynamic motion to better estimate the HJCs. However, obtaining subject-specific CT data prior to motion analysis may not be often feasible or cost-effective. The next step in the ongoing study is to introduce a corrective factor that can compensate for the discrepancy between the locations of CT markers and anatomical bone landmarks. This would provide an improved estimation of the HJCs during dynamic motions, using a more subject-specific anatomical model. The understanding of subject-specific modelling could be implemented in applications involving asymmetrical hip joint geometries, motion and loading, as well as subject-specific joint pathologies such as femoroacetabular impingement.

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Table 1: Peak hip angles and hip moments in the sagittal, frontal, and transverse planes and RMSE

Plane	Measures		Peaks		% Difference	RMSE
			Davis	CT Markers		
Sagittal	Angle (°)	Flexion	109.5	111.6	0.5	1.9
	Moment (N·mm/kg)	Descent	-36.7	-54.6	9.8	23.8
		Squat	-839.1	-847.8	0.3	
		Ascent	-84.9	-99.8	4.0	
Frontal	Angle (°)	Abduction	-10.6	-10.3	0.7	0.2
	Moment (N·mm/kg)	Descent	-142.1	-148.8	1.2	58.7
		Squat	172.4	252	9.4	
		Ascent	-187.8	-189.3	0.2	
Transverse	Angle (°)	Internal Rotation	27.7	33.4	4.7	5.5
	Moment (N·mm/kg)	Descent	19.1	35.2	14.8	19.8
		Squat	-157.3	-132.6	4.3	
		Ascent	31.7	46.5	9.5	