Effects of Skin Movement Artifacts on Kinematics and Kinetics of the Knee During Cycling

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Abstract—**Effects of soft tissue artifacts (STA) on the calculated kinematic and kinetic variables at the knee during cycling has never been reported in the literature. The current study aimed to bridge the gap. Five healthy young adults cycled on an ergometer with instrumented pedals mimicking rehabilitation conditions. The subject wore 30 skin markers on the pelvis-leg apparatus while the marker trajectories were measured using a motion capture system, and the knee was imaged at 30 Hz by a bi-plane fluoroscopy system. Joint kinematic and kinetic variables were calculated using skin markers and bone data separately, the latter being the gold standard. The results showed that using skin marker data the knee joint angles, shear forces and moments were underestimated and translations were overestimated. However, these effects were relatively small in the sagittal plane. The current results will be helpful for developing guidelines for using skin markers to study cycling motion.**

Keywords—**Soft tissue artifacts; Cycling; Knee; Joint angles; Joint moments.**

I. INTRODUCTION

Cycling has played an important role in transportation, recreation, and sport in our daily lives. Exercises with cycles have also grown in popularity over the past 10 years. About 15% of American adults and 24% of Canadian adults report cycling at least once a week for recreation or exercise purpose. However, with the increasing popularity of cycling, an increasing proportion of the billions of cyclists worldwide are also suffering from cycling-related overuse injuries. There has been a need in the development of injury-specific cycling exercise programs for the rehabilitation of the lower extremities, which requires a comprehensive knowledge of the biomechanics of cycling. Generally, biomechanics of the lower limb joints during cycling exercises has mostly been studied using skin marker-based motion analysis techniques and is subject to soft tissue artifacts (STA). However, no study has reported a complete assessment of the effects of STA on the calculated joint center motions, angles, shear forces, and moments at the knee during this activity. The current study aimed to evaluate in vivo the STA effects on these calculated variables by integrating 3D fluoroscopy and stereophotogrammetry. It was hypothesized that STA would significantly affect these calculated variables.

II. MATERIAL AND METHODS

Subjects

Twelve healthy young adults (age: 22.5±2.1 years, height: 172.5±2.1 cm, mass: 64.8±10.4 kg) participated in the current study with informed written consent, as approved by the Institutional Research Board. All subjects were free of neuromusculoskeletal dysfunction.

Experimental procedure

Each subject wore 30 skin markers on the pelvis and the right lower limb, and performed cycling on an ergometer at an average resistance of 20 Nm mimicking rehabilitation conditions. The pedals of the ergometer were instrumented with 6-component load-cells for measuring pedal reaction forces during cycling. The 3D marker trajectories were measured using a 12-camera motion capture system at a sampling rate of 120Hz (Vicon Motion Systems Ltd., UK). The knee was also imaged simultaneously at 60 Hz by a bi-plane dynamic fluoroscopy system (ALLURA XPER FD, Philips). The knees of the subjects were also CT scanned and used to construct CT-based bone models. A subject static calibration was also performed. A metronome was used to keep cycling speed at 30 RPM, which gave about 120 fluoroscopic images per cycle (approximate 3° crank angle per frame) and 240 data points for stereophotogrammetric system.

Data analysis

The subject-specific, CT-based bone models were registered to the fluoroscopy images using a volumetric model-based fluoroscopy-to-CT registration method [1], giving poses of the femur and tibia, and the knee joint center positions. The means and standard deviations of the bone pose errors were less than -0.4 (0.4) mm and -0.5° (0.3°) for all translational and all angular components, respectively [1]. During subject calibration without skin movement, bone coordinate systems were defined for the thigh and shank based on the registered poses of the femur and tibia following the ISB convention, which coincided with the segment-embedded coordinate systems. Meanwhile, the position of a skin marker relative to the associated bone coordinate system was taken as the reference for STA estimation. During movement, given the measured marker coordinate relative to the stereophotogrammetry system, the components of the STA in the bone coordinate system at time t, corresponding to the anterior/posterior (A/P), proximal/distal (P/D) and medial/lateral (M/L) components, were calculated as the current position of the marker relative to the bone and fluoroscopy coordinate systems, respectively. The gold standard positions of these markers, i.e., those of the so-called virtual bone markers (VBM), in the fluoroscopy coordinate system were obtained.

The knee joint angles were obtained following a z–x–y Cardanic rotation sequence, using both skin marker and VBM data. With the measured pedal reaction force (PRF), the moments about the knee joint center were calculated by considering the free bodies of the foot and shank using the measured motion data and PRF. The results from the VBM were taken as the gold standard. The knee joint center (KJC) was defined as the mid-point of the trans-epicondylar axis in the anatomical position, and its movement in the shank coordinate system as the knee joint center translation. Inertial properties for each body segment were obtained using an optimization method [2]. The calculated moments were normalized to body weight and leg length. Translations, forces, and moments were reported with respect to the shank embedded anatomical coordinate system.

Statistical Analysis

Descriptive statistics data was reported for maximum differences throughout analyzed cycle between the results obtained from skin marker and VBM, as well as root mean square of errors (RMSE). To compare the results from skin markers and VBM, a paired t-test was used for each time instances of the whole cycle at an increment of 3° crank angle (i.e. 120 data points) for each of the variables. A significant level of 0.05 was set for all tests.

III. Results

Accurate 3D skeletal kinematics of the knee during cycling was measured using the 3D fluoroscopy method, giving accurate joint rotations, KJC translations, forces and moments that were taken as the gold standard.

Rotations and translations

Skin markers underestimated significantly the knee flexion angles at crank angles of 0°-109° and 180°-269°(corresponding to knee flexion angles of 104.8°-35.21° and 21.92°-84.46°), the maximum difference being 8.68°(2.34°) with a RMSE of 5.02°(1.58°) (Fig. 1a, Table 1). The abduction angles were also underestimated significantly at crank angles of 0°-122° and 260°-360° (corresponding to knee flexion angles of 104.8°-25.59° and 78.91°-104.8°), the maximum difference being 7.02°(3.48°) with a RMSE of 4.42°(2.76°). Internal rotation angles were no significance found in a crank cycle.

Compared to the gold standard, the posterior translations of the KJC calculated from skin markers were significantly overestimated at crank angles of 45°-133° and 180°-269° (corresponding to knee flexion angles of 83.81°-18.62° and 21.92°-84.46°) (Fig 1b, Table 1), with a maximum difference of 8.23 (2.72) mm and a RMSE of 4.83 (2.23) mm. Distal translations of the KJC was also significantly overestimated at crank angles of 0°-121° and 146°-360° (corresponding to knee flexion angles of 104.8°-26.3° and 13.27°-104.8°), the maximum error being 14.96 (5.15) mm with a RMSE of 10.02 (4.00) mm. Lateral translations of the KJC was significantly underestimated at crank angles of 0°-29° and 282°-360° (corresponding to knee flexion angles of 104.8°-93.14° and 91.87°-104.8°), the maximum error being 7.25 (2.67) mm with a RMSE of 4.31 (2.40) mm.

Forces and moments

Knee joint forces calculated from skin markers were slightly different from the gold standard for the anterior/posterior, proximal/distal components (maximum error less than 2.97% of the maximum value of the gold standard). Significant differences were found at crank angle of 279-301° for the medial/lateral force component. Maximum difference was 8.55 (4.98) N and a RMSE of 3.51 (2.32) N (Table 1).

The extensor moments calculated using skin markers were significantly underestimated at crank angles of 0°-131° and 326°-360° (corresponding to knee flexion angles of 104.8°-19.76° and 107.8°-104.8°), the maximum difference being 2.82 (1.20) Nm with a RMSE of 1.27 (0.51) Nm. The abductor moments were significantly underestimated at crank angles of 0-44° (corresponding to knee flexion angles of 104.8°-84.46°), the maximum difference being 2.40 (1.24) Nm with a RMSE of 0.94 (0.46) Nm. In contrast to the other two components, the external rotator moments were small (maximum value: 3.35 Nm) and significant difference was found at crank angles of 106-127° (corresponding to knee flexion angles of 37.54-22.23°).

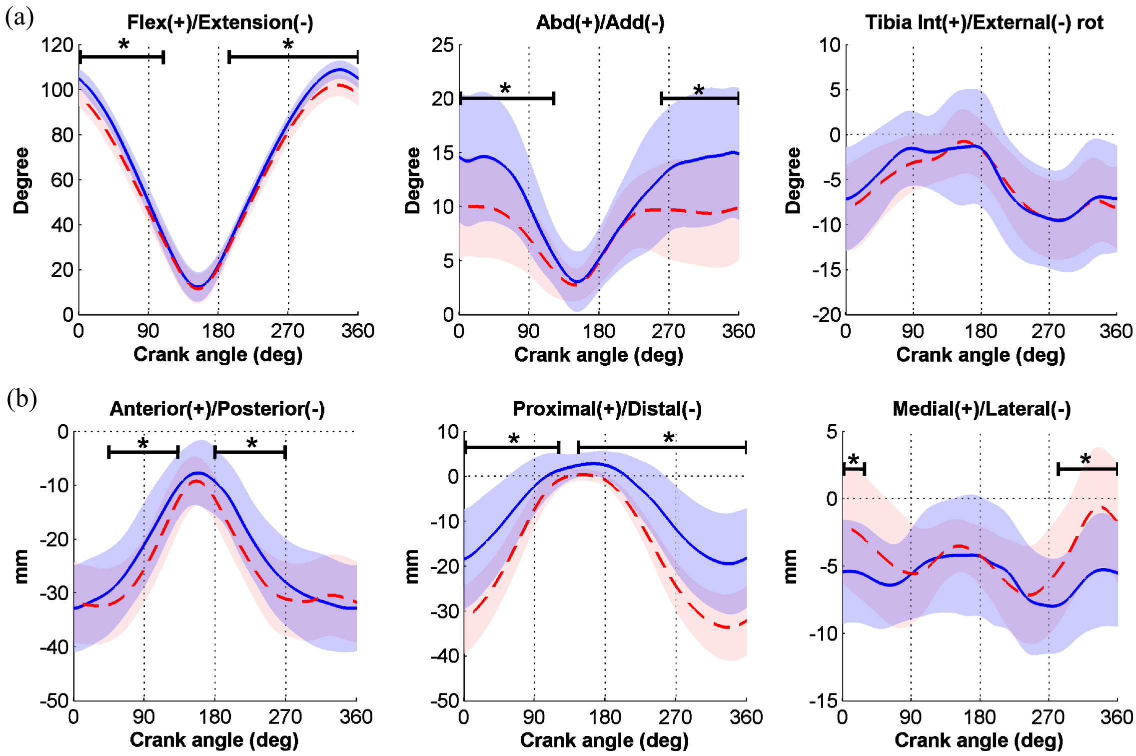

Figure 1. Mean curves of the angles (a) and joint center translations (b) of the knee obtained using skin markers (red dashed lines) and VBM (blue solid lines, gold standard) during a complete cycling cycle. Blue and red areas represent one standard deviation. Ranges marked with asterisks represent significant differences between the means.

Table 1. Means (SD) of the maximum differences and RMSE values of the calculated knee angles, translations, shear forces, and moments across all subjects, also as percentages of the maximum value obtained from VBM.

		Maximum Differences		**RMSE**	
Angles		(Degree)	(%)	(Degree)	(%)
	Flexion/Extension	8.68 (2.34)	8.00 (2.21)	5.02 (1.58)	4.63 (1.49)
	Abduction/Adduction	7.02 (3.48)	47.88 (20.29)	4.42 (2.76)	28.90 (13.54)
	Internal/External	7.39 (2.53)	75.60 (29.88)	3.77 (1.32)	38.78 (16.11)
Translations		(mm)	(%)	(mm)	(%)
	Anterior/Posterior	8.23 (2.72)	26.89 (13.15)	4.83 (2.23)	16.19 (10.23)
	Proximal/Distal	14.96 (5.15)	121.43 (149.48)	10.02 (4.00)	82.55 (102.37)
	Medial/Lateral	7.25 (2.67)	91.72 (45.20)	4.31 (2.40)	53.17 (27.52)
Forces		(N)	(%)	(N)	(%)
	Anterior/Posterior	4.00 (0.90)	2.97 (0.97)	1.54 (0.41)	1.11 (0.29)
	Proximal/Distal	2.72 (1.02)	1.69 (0.77)	1.02 (0.33)	0.62 (0.21)
	Medial/Lateral	8.55 (4.98)	37.00 (34.57)	3.51 (2.32)	15.70 (16.64)
Moments		(Nm)	(%)	(Nm)	(%)
	Abduction/Adduction	2.40 (1.24)	20.82 (17.98)	0.94 (0.46)	8.06 (6.48)
	Internal/External	1.38 (1.33)	36.34 (34.64)	0.59 (0.52)	16.16 (16.48)
	Flexion/Extension	2.82 (1.20)	6.09 (2.08)	1.27 (0.51)	2.74 (0.93)

IV. Discussion

The current study aimed to assess the STA effects on the calculated kinematics and kinetics of the normal knee during cycling. The results showed that STA caused a significant underestimation in the knee angles and moments, but mostly overestimated significantly of joint center translations at high knee flexion angles. Skin markers underestimated the knee flexion angles mainly due to posterior movement of the markers on the lateral and medial epicondyles during knee flexion. Skin markers also overestimated the distal joint center translations throughout crank cycle except for crank angles from 121°~146°, primarily because the markers on the proximal shank moved distally during knee flexion. It is noted that large standard deviations were found between subjects in most of the angular displacement components, suggesting that subject-specific STA compensation is necessary.

Joint moments are often used to assess the function of muscles during exercises. Muscle moments were generated to counteract the external moments as a product of reaction pedaling force and its leverarm lengths at the KJC, the calculated moments were affected by the KJC position. Therefore, inaccuracies in the KJC positions will lead to inaccurate joint moments. The maximum differences and RMSE in joint moments found in the current study indicate that care should be exercised when interpreting results obtained from skin markers during cycling.

Large variation between the subjects may also suggest that average patterns of the STA and the associated effects may not apply to individual subjects. Subject-specific compensation for the effects of STA is necessary, especially for the interpretation of subtle but significant differences between subject groups in clinical studies.

V. Conclusions

In conclusion, the current study reported the first data on the STA effects on the calculated knee kinematics and kinetics in healthy young subjects during cycling. The results will be helpful for the interpretation of results in future skin marker based cycling studies, and for developing STA compensation methods for future applications.

VI. Acknowledgment

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VII. References

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