ISB 2013 BRAZIL
XXIV CONGRESS OF THE INTERNATIONAL SOCIETY OF BIOMECHANICS
XV BRAZILIAN CONGRESS OF BIOMECHANICS

CONSIDERATIONS ON MARKER SOFT TISSUE ARTEFACT PROPAGATION TO BONE POSE ESTIMATES

[1] Eleni Grimpampi, [1] Valentina Camomilla, [2] Andrea Cereatti, [1] Aurelio Cappozzo
[1] Università degli Studi di Roma "Foro Italico", Roma, Italy; web: www.lablab.eu
[2] University of Sassari, Sassari, Italy; email: acereatti@uniss.it

SUMMARY
This paper deals with the propagation of the artifacts that affect each skin marker (STAs) of a cluster, as used in stereophotogrammetry, to pose, shape, and size of that cluster during movement. In particular the resulting artifacts that affect the cluster is described as made by a rigid, a deformation, and a scaling component. Quantitative information about these three artefact components were provided by a Procrustes analysis. This objective was accomplished by using photogrammetric data relative to both markers positioned on the skin and on hip-bone and femur pins during *ex-vivo* passive knee flexion-extensions. The analysis was carried out for all possible combinations of four skin markers, out of twelve placed on the thigh. Particularly, the rigid displacement component exhibits maximal values between 5-19 mm for cluster centroid displacement, 1.1-15.1 deg for orientation vector variation; while cluster deformation exhibited values between 0.93-8.47 mm and scaling ranged between 0.4 and 8.2 %. Results showed that bone-pose estimators accounting only for cluster deformation do not assure for accurate bone pose estimates. Thus, future work is needed to develop new pose estimators that include modelled information on the real STA.

INTRODUCTION
The estimate of the instantaneous pose (position and orientation: BP^G) of a bone, relative to an inertial observer (global frame: *G*), during the execution of a physical exercise and using data provided by non-invasive experimental techniques, is a central and yet inadequately solved problem in human movement analysis [1].
The assumption of rigidity of the bone and the consequent use of a bone-embedded frame (*B*) to numerically describe its pose is commonly accepted in the present context. The BP^G is most commonly obtained indirectly using the positions, relative to *G*, of at least three non-aligned physical markers (marker cluster) located on the surface of the relevant body segment under analysis. These positions are reconstructed using stereophotogrammetry. Relevant marker position data are provided as input to a mathematical model (bone pose estimator) the output of which is the instantaneous global pose (CP^G) of a frame (*C*) rigidly associated with the cluster. Through *ad hoc* calibration procedures, the pose of *C* with respect to *B* (CP^B) is obtained [2]. By cause of the unavoidable deformation of the soft tissues, the markers move with respect to *B*. This movement is to be regarded as an artefact and, given its origin, it is referred to as marker soft tissue artefact (STA). As a consequence, the pose estimator defines a frame *C* which is not rigidly connected with *B*, as it would be in the absence of STAs.
The STA may be formally defined as the displacement of a marker relative to a given reference position, as observed in *B*. Based on a mathematical approach known as Procrustes analysis [3], if a marker is included in a cluster its displacement can be partly explained as the effect of simple geometrical transformations applied to the whole cluster. Namely, its displacement may be imagined to be the sum of three independent components:
1) a component which is the result only of a rigid movement of the whole cluster relative to *B* (*rigid displacement* component);
2) a component which is caused only by a change in shape of the cluster (*deformation* component).
3) a third component which is caused only by a change in size of the cluster (*scaling* component);
A pose estimator should pursue the minimization of the propagation of the STAs to the estimate of BP^G, that is, minimizing the variations of (CP^B) during the analyzed movement.
Several published studies provide information concerning individual marker's STAs, mostly with reference to the human thigh and shank [4,5,6], and their effects on the estimate of knee joint kinematics [7,8,9,10]. Conversely, few investigations are available regarding the above-illustrated components though without providing quantitative information about STA amplitude [11] or conducting experiments not on humans [12].
The aim of this study is to provide quantitative information of the STA components conducting experiments in humans.

METHODS
Three intact adult male cadavers were used for the study. All subjects had no history of injuries on their lower extremities. Steel pins were inserted into the right tibia, femur, and pelvis. Each pin was equipped with a four marker cluster (minimal distance between two markers: 70 mm). In addition, twelve skin markers were placed on the right thigh along three longitudinal lines: antero-medial, anterior and antero-lateral line. A photogrammetric system (Vicon MX, 120 frames/s) was used for the reconstruction of skin and

pin marker trajectories in *G* while an operator made the hip and knee flex and then extend (mean range of movement: 55 and 119 deg for hip and knee, respectively). Three trials for each subject were performed. The pelvic and femoral anatomical landmarks relative to the corresponding local frame were determined through anatomical calibration, using a pointer equipped with a cluster of four markers [2]. Moreover, the hip joint center was estimated using a functional approach [13]. Then, the reconstructed position of each skin marker was represented in the femoral frame. All possible combinations of four skin markers, out of the 12 available, were used to generate 495 skin marker clusters. The clusters with insufficient 2D isotropy index were discarded and 332, 416 and 370 clusters were analysed for subject 1, 2, and 3, respectively. Given the reconstructed position of four markers forming a cluster, the orientation of the axes frame *C* was determined using a geometric rule.

A cluster template was defined as the mean of the sampled configurations (or "mean shape") of the cluster (Dryden and Mardia, 2002). The relevant reference frame, representing the pose of the template, was defined with respect to *B* using a geometric rule. The cluster template was, thereafter, superimposed to the current marker cluster by optimally translating, rotating and uniformly scaling it. To this purpose the root mean square distance (RMSD) between the corresponding points of the two clusters was minimized [3].

The STA components were described as follows: for the *rigid displacement* component, the pose of *C* with respect to *B* was reconstructed during the analyzed movement using Procrustes analysis and a least squares approach for the registration [12], its variation with respect to a reference pose assumed at time zero was described as the maximal displacement of the cluster centroid (p_c^B) and the maximal change in the orientation vector (ϑ_c^B). The *deformation* component was quantified as the mean of the point to point distances between model-cluster and time varying skin marker-cluster (δ_c). The *scaling* component was defined as the ratio of the size of the measured marker cluster and the size of the cluster-template, computing sizes as the root mean squared distance from the markers to their centroid (k_c). Moreover, Pearson's correlation coefficient was used to assess the relationships between these parameters.

RESULTS AND DISCUSSION

Results confirmed previous evidence of a residual rigid motion effect of skin marker artefact using a pose estimator based on the above mentioned approach. No correlation was found between the parameters for the three subjects. Moreover, all the parameters used to describe the STA components exhibited a large variability across subject and cluster selection. The marker cluster undergoes simultaneously to a rotation and a translation with respect to the bone and a size and shape variation during the specific task tested. Particularly, the rigid displacement component exhibits values between 5-19 mm for p_c^B and values between 1.1-15.1 deg for ϑ_c^B, while δ_c exhibits values between 0.93-8.47 mm and k_c between 0.4-8.2 %.

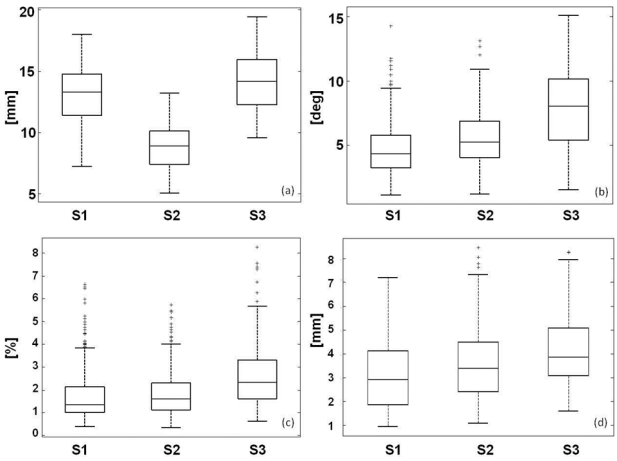

Figure 1: Box-plots of: (a) p_c^B, (b) θ_c^B, (c) k_c, and (d) δ_c of CP^B relative all not quasi-collinear clusters for the three trials of the three subjects S1, S2, S3. Outliers are also shown.

The aim of this study was to assess the various components (rigid, deformation, and scaling) of the STA effect on the estimate of femoral bone pose BP^G at the marker cluster level, using the Procrustes method tested with an *ex-vivo* approach. Firstly, a number of limitations need to be considered in association with the present study: 1. a limited number of cadavers was available, not allowing for a statistical description of the STA characteristics; 2. the experimental method used. Taking into consideration that no active muscular contraction exists, the main cause of the described STA was the skin stretching related to the knee/hip movement during the task. Despite the above-mentioned limitations, the STAs observed in this study had similar amplitude to that of previous *in-vivo* studies involving tasks with a similar range of motion and marker location [9,14].

CONCLUSIONS

The findings of this study suggest that accounting only for the cluster deformation does not guarantee for accurate bone pose estimate, thus calling for the development of new approaches.

REFERENCES

1. Cappozzo A, et al., *Gait Posture*. **21**:186-196, 2005.
2. Cappozzo A, et al., *Clin Biomech*. **10**:171-178, 1995.
3. Dryden IL, et al., *New York: John Wiley & Sons*, 2002.
4. Garling EH, et al., *J Biomechanics*. **40**:S18-S24, 2007.
5. Alexander EJ, et al., *J Biomechanics*. **34**:355-361, 2001.
6. Peters A, et al., *Gait Posture*. **31**(1):1-8, 2010.
7. Ryu T, et al., *Hindawi Publishing Corporation, Anatomy Research International*, **2012**, Article ID 123713, 2012.
8. Benoit DL, et al., *Gait Posture*. **24**:152-164, 2006.
9. Akbarshahi M, et al., *J Biomechanics*. **43**:1292-1301, 2010.
10. Tsai T-Y, et al., *J Biomechanics*. **44**:1182-1188, 2011.
11. Andersen MS, et al., *Gait Posture*. **35**(4):606-11, 2012.
12. Taylor WR, et al., *J Orthop Res*. **23**(4):726-34, 2005.
13. Cereatti A, et al., *J Biomechanics*. **42**:818-823, 2009.
14. Gao B, et al., *J Biomechanics*. **41**(15):3189-3195, 2008.