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HUMAN KINEMATICS RECONSTRUCTION FROM MARKERLESS AND MARKER-BASED MOTION ANALYSIS SYSTEMS BY MODEL-BASED APPROACH

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INTRODUCTION

Clinically relevant muscle information (length, moment arms) is difficult to estimate directly in clinical settings. Quality of the obtained muscle approximation strongly relies on the underlying bone and kinematic data used to create the joint models that will be crossed by the muscle spatial path. We present a method that allows fusing accurate joint kinematic information with relatively crude motion analysis (MA) data collected using either marker-based stereophotogrammetry (i.e. bone displacement collected from reflective markers glued on the subject's skin) or markerless single-camera hardware. The obtained kinematical model can then be used for further modeling (e.g. muscle moment arm or excursion by addition of relevant data) which quality will depend on the underlying kinematic model. Typically, a global optimization method based on mechanical modeling could be applied to adjust model parameters to a particular motion. Different sets of joint constraints related to joint kinematics (e.g. joint surface geometry, ligament information and joint mechanism) were previously implemented in order to assess their influence on the lower limb kinematics during gait [1]. This approach requests implementation of collision detection and reaction mechanism procedures such as the ones available from most commercial multibody dynamics software.

This abstract describes a novel model-based approach (MBA) [2] for human motion data reconstruction using scalable method that combines joint physiological kinematics [3,4,5] with limb segment poses. This approach is an extension of a previously published double-step registration method [6], developed for lower limbs MA. Advanced computer graphics visualization and user interface allow displaying fusion results and measurement graphs simultaneously. These data can then be processed during further statistical analysis.

METHODS

Generic morphological bone models (GM) for the lower and upper limbs (LL and UpL, respectively) were collected during past European-funded projects (VAKHUM, LHDL and DHErgo) from fresh-frozen cadaveric specimens obtained from the ULB Body Donation program using medical imaging [4] (Figure 1).

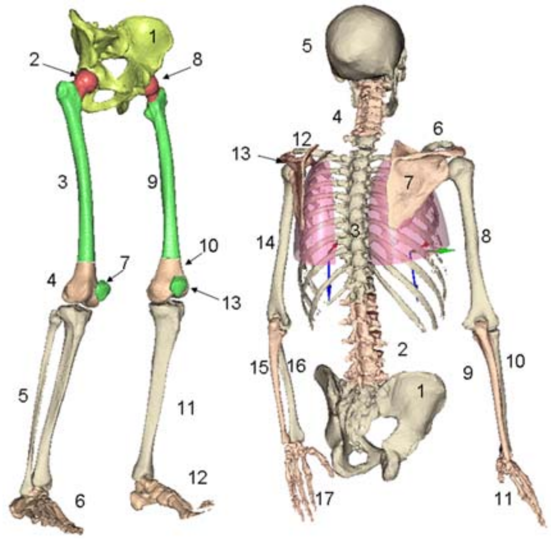

Figure 1. Generic models and segment numbering used in this study. A total of 30 links and 144 DoFs are present in the model (LL: 13 links, 72 DoFs; UpL: 17 links, 72 DoFs). Note that each femoral bone was divided into three parts (head, diaphysis and distal epiphysis) to allow customized reorientation of these components and speed up optimization when required. Ellipsoid surfaces are included on both side of the thorax to constrain scapular gliding using ellipsoids principal axes.

Twelve of these specimens were used to collect in-vitro LL joint kinematics data for the hip, knee and ankle joints using 6DoFs instrumented spatial linkage [3]. In-vitro joint kinematic data related to the shoulder complex were collected on 2 fresh-frozen specimens. In-vivo motion data related to the shoulder complex were obtained from 3 volunteers with TF clusters glued on each segment-of-interest which ALs were previously manually palpated [7]. Motion data were collected along each anatomical planes (passively and actively for the specimens and volunteers, respectively). For UpL bones, the projections of each DOF related to the clavicle, scapula and humerus pose vectors on the thorax anatomical frame were plotted. In total 144 (2 proximal bones, 2 linear and parabolic fitting, 6 DoFs proximal and 6 DoFs humerus bones) plots were processed by linear and parabolic fitting. Then shoulder pose prediction was implemented following the parabolic weighted multiple regression. This approach allowed predicting the 6 DoFs-dependent motion of the clavicle and scapula from the combination of up to 6 DoFs humerus behavior relative to the thorax.

The method presented in this paper has been developed in order to be used with standard marker-based stereophotogrammetry (MBS) systems (Vicon MXT40S) or single camera markerless system (MLS) (Microsoft Kinect™). Obviously, these two systems show different specifications and qualities. MBS systems are relatively accurate and allow real 3D analysis. However, they are relatively costly and time-consuming due to the manual placement of markers being analyzed. MLS on the other hand are less accurate, but are cheap and allow quick data collection. Integrating both systems into one unique modeling pipeline would allow collect data according to the user requirements and equipment availability. In-vivo MBS data were collected on 7 volunteers. Subjects were asked to perform various movements (walking, sitting on a chair, squatting, jumping, large free motion of the upper limbs, etc). On two volunteers, MBS and MLS data were collected simultaneously for comparison.

Both MBS and MLS joint 3D pose data were processed frame by frame using a model-based approach (MBA). The MBA human skeleton (Figure 1) included two tree-structured parts (LL and UpL) both starting from the pelvis. This allows organizing input data processing according to the available motion data: UpL motion only, LL motion only or simultaneous UpL and LL motions (i.e., during full body analysis) MBA model registration to the captured motion data, either MBS or MLS, was based on an inverse kinematics (IK) approach. The purpose of the IK step is to find the set of generalized coordinates (joint angles and positions) for the model that best fits the motion data recorded for a particular subject. The IK processed each time step (i.e. motion frame) available from the motion dataset and computes generalized coordinate values which position the model in a pose that fits input AL coordinates for that particular time step.

RESULTS AND DISCUSSION

LL joint kinematics (i.e., angular velocity and accelerations) comparison (MBS vs MLS) was evaluated for squat motion using Grood and Suntay convention versus joint flexion/extension motion. The range and shape of the estimated accelerations plots are similar for both kinds of input data. Both data sets show some hysteresis behavior of accelerations, which could be explained by gravity factor. The hip joint motion amplitude showed a 30° difference between both input systems. This was due to the nature of the deep squatting movement, requesting a large hip and knee flexion, which led to a poor visibility of the hip area by the single camera MLS.

Glenohumeral joint kinematics comparison between results obtained from MLS and MBS data was evaluated for both humeral bones. Humeral elevation are given in different planes corresponded an arbitrary motion of the humerus. The glenohumeral joint center translations were obtained relative to the thorax local CS. Mean differences for the plots are presented in Table 1.

Table 1. Mean error (MLS vs MBS data) of the both humerus motion. R = right; L = left; El = humeral elevation; X = anterio/posterior and Y = inferio/superior translations; dist = distance from sterno-clavicle to humerus head center.

1.	2.	3.	4.	5.	6.	7.	8.
R_El (°)	**L_El (°)**	**R_X (mm)**	**R_Y (mm)**	**L_X (mm)**	**L_Y (mm)**	**R_dist (mm)**	**L_dist (mm)**
9.7	**5.2**	**5.6**	**6.1**	**3.1**	**2.7**	**1.9**	**1.0**

The presented results and kinematics analysis show that MBS and MLS model-based reconstructions lead to physiologically-correct human kinematics. The results are in agreement with the literature, for example the shoulder rhythm is respected and the knee special mechanism and patella displacement are included in the model [8,9]. Furthermore, MLS data seems to lead to results comparable to MBS data. MLS data collection could then be an interesting alternative for collecting data in settings where a cumbersome MBS system is difficult to use (for example, at a patient's home). The method presented in this paper has been implemented for motion data collected from conventional MBS systems and MLS cameras. The accurate underlying generic data and the presented method allow estimating biomechanically relevant information (e.g. physiologically correct joint kinematics and muscle line-of-actions behavior). Fusion results are, however, sensitive to the original accuracy of both morphological and kinematic data. Artifacts may arise due to the soft-tissue deformation during the data collection performed on the subject undergoing the clinical data collection.

CONCLUSIONS

The overall (MLS vs. MBS) method showed satisfactory accuracy and therefore the proposed system method is available for further exploitation of the underlying model. Further studies, using the presented methods, should improve the ability to interpret musculoskeletal mechanism in biomechanical and clinical research, for example by analyzing the muscle behavior (i.e., instantaneous muscle length and moment arms) in different population of subjects (normal and pathological).

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