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Quantification of gait parameters with inertial sensors and inverse kinematics

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ABSTRACT

Measuring human gait is important in medicine to obtain outcome parameter for therapy, for instance in Parkinson's disease. Recently, small inertial sensors became available which allow for the registration of limb-position outside of the limited space of gait laboratories. The computation of gait parameters based on such recordings has been the subject of many scientific papers. We want to add to this knowledge by presenting a 4-segment leg model which is based on inverse kinematic and Kalman filtering of data from inertial sensors. To evaluate the model, data from four leg segments (shanks and thighs) were recorded synchronously with accelerometers and gyroscopes and a 3D motion capture system while subjects (n = 12) walked at three different velocities on a treadmill. Angular position of leg segments was computed from accelerometers and gyroscopes by Kalman filtering and compared to data from the motion capture system. The four-segment leg model takes the stance foot as a pivotal point and computes the position of the remaining segments as a kinematic chain (inverse kinematics). Second, we evaluated the contribution of pelvic movements to the model and evaluated a five segment model (shanks, thighs and pelvis) against ground-truth data from the motion capture system and the path of the treadmill.

Results: We found the precision of the Kalman filtered angular position is in the range of 2–6° (RMS error). The 4-segment leg model computed stride length and length of gait path with a constant undershoot of 3% for slow and 7% for fast gait. The integration of a 5th segment (pelvis) into the model increased its precision. The advantages of this model and ideas for further improvements are discussed.

1. Introduction

The implementation of small lightweight inertial sensors has prompted many research groups to investigate their usefulness for recording of human gait. These projects may aim at answering specific questions and therefore follow different approaches: for long-term activity monitoring a single sensor attached to the waist may reliably record the number of steps but lacks precise information about step size (Del Din et al., 2016; Trojaniello et al., 2014). For analyzing gait in patients with spasticity or Parkinson's disease limb angles and step sizes are crucial and this requirement necessitates sensors on each limb segment (Curtze et al., 2015). Data from such equipment can be analyzed by dedicated algorithms to detect important events (toe-off, heel-strike) as well as to compute stride length by integrating the data from gyroscopes mounted on the shank (Aminian et al., 2002; Tong and Granat, 1999). Also, the signal of a shoe-mounted accelerometer can be integrated twice, to obtain step length during the swing phase (Sabatini et al., 2005). Three-dimensional displacements of a body part during cyclical motions can also be reconstructed by use of accelerometers and applying Fourier-analyses (Sabatini et al., 2015). Gyroscope signals can be integrated into this process to determine the angular position of the sensor and to compute the direction of the acceleration vector (Alvarez et al., 2007; Rebula et al., 2013). With different filters (least mean squares, recursive least squares and Kalman), signals of accelerometers and gyroscopes can be de-drifted (Rebula et al., 2013) or fused to obtain an estimate of the angular position of the limb without the error introduced by gyroscope drift (Bennett et al., 2014; Olivares et al., 2016). The Kalman filter is a common tool to obtain accurate estimates of orientation from inertial sensors. In the present paper, we are reporting the application of Kalman filtering of accelerometer and gyroscope

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data for accurate limb angle estimation and second, the integration of such data into a leg model realized as a kinematic chain. A similar model which was based on the integration of gyroscope data had been presented previously (Salarian et al., 2004). In our model, the stance foot is taken as a pivotal point from which the positions of four leg segments (shanks and thighs) are calculated. With this method, the trajectory of the swing foot can be computed. Third, pelvic movements were studied to assess their contribution to the forward movement of the body during gait and lastly, a 5-segment model including shanks, thigh and pelvis was evaluated.

2. Methods

2.1. Inertial measurement system

We used a research prototype measurement and recording system with a microprocessor (ATXMEGA 128, Atmel, San Jose, CA, USA) with 16 analog-digital converters (12-bit) connected to an SD-card for data storage. For the recording of limb movements we used one analog gyroscope (IDG500, Invensense, Sunnyvale, CA, USA) and two analog accelerometers (ADXL335, Analog devices, Norwood, MA, U.S.A) on each shank and each thigh. Data from the sensors were collected at 200 Hz. The sensitivity was 6 levels/deg/s for gyroscopes and 2095 levels/g for accelerometers. Data were sampled with 16 times the sampling frequency to enhance the resolution of the system (oversampling). Sensors were connected with the processor by dedicated shielded sensor cables. The sensors were secured with elastic bandages on the middle of the shin bones and on the middle of the quadriceps muscle of both legs. With a ruler, we controlled that the ground plate of the sensors was aligned with the frontal plane of the subject. Thus, the accelerometers signaled acceleration in the direction of the vertical (z) and anterior (x) body axis and the gyroscopes sensed rotation in the sagittal plane. Sensors had been carefully calibrated. The calibration process was applied to transform digital raw values into meaningful physical unites (deg/s and m/s^2), as well as to compensate for other sources of error such as non-orthogonalities and Angle Random Walk (Camps et al., 2009). For the accelerometer, we applied an ellipsoid fitting algorithm (Camps et al., 2009). For the gyroscope, we employed a method directly based on Ferraris et al. procedure (Ferraris et al., 1995) by rotating the device around a known angle (180°). Multiple rotations were carried out and the computed scale factors were averaged. Bias was computed by leaving the device static for a few seconds and computing the mode to compensate for noise.

2.2. Motion capture system

We used an Oqus camera system (Qualisys AB, Gothenburg Sweden). Three infrared reflecting spheres (diameter 2 cm) were attached to each shank and each thigh. On each limb, 2 spheres were attached to the anterior portion of the limb at a distance of 20 cm and one at the back side of the limb at the middle of the limb. Thus the reflecting markers of each segment formed a triangle in the sagittal plane of the subject. Two additional spheres were positioned at the height of the pelvic crest at the lateral aspects of the hips and one sphere was attached at the same height on the sacrum. These 3 spheres formed a triangle in the horizontal plane and allowed for the registration of hip rotations in 3D. The position of the spheres was recorded at 200 Hz by eight cameras. Resolution of the system was below 1 mm. Data were evaluated by Matlab scripts. Motion capture data were synchronized with inertial sensor data by computing the angular velocity of one shank from the motion capture system and cross-correlating this trace with the gyroscope trace of this shank (MATLAB function 'xcorr').

2.3. Evaluation

All analyses were performed off-line by dedicated Matlab scripts (Fig. 1). Inertial sensor's data from each segment (two accelerometers and one gyroscope) were fused by Kalman filtering to obtain the segment's angular position at each sampling interval (Fig. 2). In this case, we applied a two-state (Euler angle and gyroscope bias) standard Kalman Filter (Olivares et al., 2016). Limb angles in the sagittal plane computed by the Kalman filter were then compared to the data obtained from the motion capture system which served as 'gold standard'. For this evaluation, only the middle 50% of the traces were used because they represented gait data obtained during constant gait velocity. The difference of these two data sets were described by computing the root mean square error (RMSE) for each pair of traces. To see whether the error depends on gait velocity, this was done for three gait velocities. To evaluate gait parameters, gait events were detected and labelled by software and confirmed or corrected after visual inspection: first, the peak of the gyroscope trace during the swing-phase was detected (Matlab routine 'findpeak'). Then, the heel-strike moment was determined by detecting the trough of the shank gyroscope trace after this peak (Fig. 3). The toe-off-moment was computed according to our previous work in which we showed that this moment is slightly after the trough of the gyroscope trace which precedes the swing-phase peak (Bötzel et al., 2016; Bötzel et al., 2017). The heel-strike determined the stance foot which was taken as a pivotal point for computing the trajectory of the following step. These data allowed for the determination of stance phase duration, swing phase duration, as well as initial and terminal double support and cadence. Gait cycle duration was updated at the heel-strike of the right foot. Sequentially, the position of the knee of the stance limb, hip joint, opposite knee and swing foot (2D data, in the sagittal plane) were computed by trigonometric equations on the basis of the angular position of the limb and the length of each individual segment (Flow chart of computations: Fig. 1, inverse kinematics, Fig. 2). The trace of the swing foot allowed for the calculation of foot clearance and step length. Gait velocity was updated after each step. Gait path was calculated by summing up step length. Hip movements were not included in this first model. The model was evaluated by comparing gait path length calculated by the model with the gait path determined by the motion capture system and the gait path of the treadmill. For a second model, pelvic rotations measured by the motion capture system in three dimensions were analyzed to obtain an estimate of the contributions of the pelvis to the forward motion of the subject. These data (rotations in 2 axes) were integrated into the sensor-based model by including a 5th segment (pelvis). The length of this segment was computed by subtracting twice the distance between trochanter and hip joint center (femur length x 0.2 (Rauber and Kopsch, 1906)) from the distance of the two pelvic motion capture spheres. These data (step length and gait path length) were compared to data obtained from the motion capture system to which the positional signal of the treadmill had been added and also compared to the length of the gait path which was given from the treadmill.

2.4. Subjects and gait task

Twelve healthy men (age 25–40), who were recruited from hospital staff, walked on a treadmill (Woodway Inc., Waukesha, WI, USA). For each of three velocities (2, 4, and 6 km/h), one recording, lasting about 1.4 min each, was performed. The length of the thigh was measured from the greater trochanter to the space between lateral condyle and tibia, and shank length was measured from this point to the floor (thus including soles). The length of the thigh was corrected by adding 8% of the measured shank length to account