IMPACT OF DYNAMIC CALIBRATIONS ON MEASURMENTS OF A WEARABLE MOTION ANALYSIS SYSTEM

Andrew Sexton, Chris McGibbon, Adam Wilson, Glen Hughes, Bernie Hudgins
Institute of Biomedical Engineering, University of New Brunswick, Fredericton, New Brunswick, Canada

INTRODUCTION

There is considerable interest in developing human motion capture technologies that can be used outside the clinic or laboratory environment. Such technologies would enable motion-related clinical measures and rehabilitation outcomes measures to be acquired in the patients' natural environments: e.g. at home, at work, or in the community. The ability to acquire motion-related measures in patients' real-world environments (at any time of day, for any length of time) would provide rehabilitation professionals with data that better reflects patients' real-world functional limitations. To address this need, we have developed a wearable motion capture system for the human arm, to enable measurement of arm kinematics during rehabilitation exercises and daily activities (eating, dressing, etc.) in any environment. We plan to use this device for acquiring arm function measurements of stroke patients.

There are approximately 300,000 Canadians currently living with the debilitating effects of stroke [1]. With an annual cost exceeding $2.7 billion, with $27.5 thousand per year per patient for acute care costs, there is a considerable challenge for rehabilitation professionals to provide efficacious treatments to restore function and reduce disability caused by stroke. Motor dysfunction experienced by stroke survivors can be due to any number of different functional syndromes such as paresis, ataxia, apraxia, visuo-perceptual deficits, or deafferentation [2]. A common functional deficit experienced by many stroke survivors is severe arm paresis. This can be characterized in general as loss of elbow and shoulder mobility caused by spasticity of flexor muscles, resulting in severe difficulties in self-care, such as feeding and dressing, and inability to perform common functional tasks, such as grasping and moving objects. Recent evidence suggests that arm motor function at 1 month post-stroke is one of the biggest predictors of stroke recovery [3]. It is therefore imperative that close monitoring of arm function in the weeks following stroke be clinically viable. Unfortunately, high costs and other burdens associated with frequent visits to the clinic or rehabilitation hospital often prevent adequate temporal resolution of arm mobility recovery assessments.

The project described in this paper seeks to develop a solution to this common clinical problem, by developing, testing and implementing a wearable motion sensor device for capturing arm motion at any desired temporal resolution, and with accuracy and reliability exceeding that of commonly used clinical instruments.

DESCRIPTION

The work performed for this paper is a continuation of the work started in [4] with the specific purpose of further validating the elbow sensor system and joint modeling algorithms, hereafter referred to as the IBME Sensor System and IBME Joint Model, when mounted on human subjects performing dynamic movements. The validation of this system was accomplished by comparing elbow joint flexion/extension angles of the IBME Sensor System to corresponding angles calculated from the measurement of a pair of arrays of reflective spherical markers using a Vicon M-Cam motion analysis system, hereafter referred to as the Vicon Marker Array System.

In general, two comparisons of the IBME Sensor System and Vicon Marker Array System are made. The first is a comparison of just the sensor readings (Sensor-to-Sensor). The second is a comparison of the output of the IBME Joint Model for the two different sensor system's inputs (Model-to-Model).

<u>IBME Joint Model</u>

Consider a joint to consist of 2 rigid segments where each segment's long axis has 2 endpoints and 1 orientation. The IBME Joint Model is a set of transforms that converts 2 6DOF sensor measurements into the 4 segment axis endpoints and 2 segment orientations. One 6DOF sensor's measurement corresponds to the upper segment of the joint and the second 6DOF sensor to the lower segment. The 6 transform values are calculated using both static and dynamic calibration movement measurements. The static calibration yields an intermediate set of transform values and the dynamic calibration refines these transform values to achieve a minimized joint gap (RMS of displacement) over the dynamic calibration data trial.

For the purposes of this paper, the angle of the joint is determined by taking the difference in orientation of the upper and lower segments of the joint model, expressed as a rotation matrix. This difference can then be expressed in clinical angles of flexion/extension, abduction/adduction and internal/external rotation using a Cardan 3-1-2 decomposition [5]. We focus on the elbow flexion/extension angle in the dynamic movements in the test protocol of this paper. The purpose of employing a joint model is to attempt to measure skeletal position and orientations as opposed to points on the surface of the body, thereby producing more clinically accurate measures of joint position, orientation and movement.

Sensor Mounting and Data Acquisition

The IBME Sensor System was very similar to that used in [4] but was now mounted on a human arm and torso (back) and the ShapeTape [6] ribbon sensor was placed in a flexible rubber sheath as seen in Figure 1.

Figure 1: Sensor Mounting

The casing of the ShapeTape and Microstrain 3DM-GX1 inertial sensor [7] was securely velcroed to a worn vest, vertically in line with the spinal column of each subject. Stretch fabric cuffs were fitted at the biceps and wrist locations of the subject and the ShapeTape then passed over the shoulder and down the arm to the wrist, being affixed with velcro to the cuffs.

The Vicon marker array plates were also attached to the subject at the same locations of the upper and lower arm sensors of the ShapeTape. The Marker plates were carefully placed over top of and in line with the ShapeTape sheath and the combination of sensors was secured in place with Velcro straps to the cuffs. The Vicon marker array plates consisted of four reflective spherical markers arranged at the four corners of a rectangular thin rigid plate with marker spacing dimensions of 70mm x 40mm. A fifth marker was placed in the upper plate simply to distinguish it from the lower plate in the Vicon capture software. The position data recorded by the Vicon capture system of these four markers can be converted to an average position and orientation of the marker array plate, thereby producing a 6DOF measure for each marker plate.

A single computer acquired both the Vicon data and the IBME sensor data simultaneously. The Vicon software and hardware allowed for an external triggering interface which clocked the sampling of the IBME sensors and synchronized the data acquisition of the two systems.

Testing Protocol

The experiments consisted of measuring elbow flexion/extension joint angle of four subjects in a seated position. The upper arm movement was static with no shoulder flexion/extension at three nominal angles of abduction, 0, 30 and 60 degrees. The lower arm performed repeated flexion/extension movements from full extension to approximately 90 degrees of flexion, typically 2 to 3 cycles per 5 second trial. The forearm rotation was kept constant at 90 degrees internal rotation, with the palm of the hand facing toward the body. Each of the 3 shoulder abduction angle positions was repeated 3 times for a total of nine randomized trials per subject.

The performance of the sensor and modeling systems was assessed by computing the absolute average difference between elbow angle measurements for the two systems, and the repeatability of these difference measures between trials and at different shoulder positions. The difference in elbow angle between systems was assessed by comparing the IBME Sensor System and the Vicon Marker Array System both before (sensor-to-sensor) and after (model-to-model) joint modeling. For each trial of each subject, the average absolute difference (AAD) of the flexion/extension angle was calculated over the 5 second trial. This average absolute error was then analyzed over all the trials and subjects. The repeatability of the differences was assessed with the intraclass correlation coefficient (ICC, Shrout & Fleiss model 2, two-way mixed). The target repeatability coefficient was ICC>.70.

RESULTS

Four subjects were recruited and provided written consent to participate in the study. The following table summarizes their data.

Table 1: Subject data

	Subject			
	1	**2**	**3**	**4**
Gender	M	M	M	M
Age	35	30	28	37
Upper arm length (mm)	270	340	350	305
Forearm length (mm)	240	325	310	270
Biceps circumference	315	375	370	335
Wrist thickness	75	80	80	80

For each of the 9 trials of each of the 4 subjects (36 trials) the AAD between the IBME Sensor System and the Vicon Marker Array System was calculated for both Sensor-to-Sensor and Model-to-Model comparisons. The average and standard deviation of AAD for all these trials is listed in Table 2. The third column lists the average and standard deviation of the difference between the Sensor and Model AAD's of each trial.

Average Absolute Difference

Table 2: Statistics of AAD for N = 36 trials

	Sensor-Sensor	Model-Model	Diff.
Average	5.53	6.09	-0.57
Std. Dev.	1.36	2.23	2.20

To test whether the difference in the Sensor-to-Sensor comparisons is significantly different from the Model-to-Model comparisons, a paired t-test was employed. Table 3 shows these results. The mean difference in AAD between comparisons was enclosed within the 95% confidence interval, and thus may be considered non-significant. It can therefore be concluded that the model does not significantly increase the uncertainties that propagate through the model equations from the sensor readings.

Table 3: Paired t-test sensor error to model error

		95 % CI			
Mean diff	**Std error**	**Lower**	**Upper**	**t**	**p**
-0.57	0.3665	-1.3116	0.1766	-1.548	0.131

Repeatability

Repeatability was assessed to determine if the sensor-to-sensor comparisons, and the model-to-model comparisons, remain consistent over repeated trials and trials at different global arm position (e.g. shoulder abducted at different angles). Data in Table 4 show these results.

We found that the trial repeatability, at same shoulder positions, was very high for both the sensor-to-sensor comparison (ICC=.949, 95% CI .718-.997, p=.002) and the model-to-model comparisons (ICC=.984, 95% CI=.913-.999, p=.001). Repeatability was also high for tests at different shoulder positions for both the sensor-to-sensor comparisons (ICC=.866, 95% CI=.299-.990, p=.002) and the model-to-model comparisons (ICC=.888, 95% CI=.415-.992, p=.01). The considerably wider confidence intervals for shoulder position ICC coefficients suggest that care must be taken when mounting the apparatus to avoid cross-talk between joint angle measurements.

Table 4: Repeatability analysis

		95% CI			
	ICC	**Lower**	**Upper**	**F**	**p**
Sensor-to-sensor					
Arm position	0.866	0.299	0.990	11.02	0.008
Trials	0.949	0.718	0.997	18.38	0.002
Model-to-model					
Arm position	0.888	0.415	0.992	9.92	0.010
Trials	0.984	0.913	0.999	74.6	0.001

DISCUSSION

The validation results presented in this paper are quite promising. By comparing the IBME Sensor System and Vicon Marker Array Sensor System it is evident that the IBME system provides consistent and repeatable measurement results of elbow joint flexion angle. The Sensor-to-Sensor and Model-to-Model comparisons provide very similar results with no statistically significance difference, indicating that the model performs as expected and does not introduce uncertainties into the measurement.

The performance of the sensor system and modeling techniques is robust for different subjects with differing shapes and lengths of arms. It is also robust over different positions of the shoulder abduction angle. However, the repeatability of measures across arm abduction positions appeared to suffer slightly, most likely due to the ShapeTape sensor binding or moving with increased arm abduction. This probably contributes to the wide confidence interval on the ICC in Table 4. Different mounting strategies may be required to reduce this artifact.

In general, the largest differences in measure were observed for extremes of elbow joint angle and, again, this may be due to mounting issues that cause the ShapeTape to buckle and twist severely around the elbow. This may be alleviated by finding better ways to mount the ShapeTape, in particular the casing, to the body. As well, modifications to our modeling techniques and calibration positions/movements to characterize these errors may be investigated. The high repeatability of the measures suggests that modeling and calibration may have significant positive effects on increasing measurement accuracy.

Much care was taken in donning the instrument to ensure sensor alignment was as close as possible to the skeletal system. This may potentially be alleviated by better calibration techniques with special positions and movements to remove sensor errors due to misalignment and mounting issues.

CONCLUSION

Overall, the IBME Sensor System seems to provide good accuracy in measurement with high repeatability across subjects. The system seems adequately capable of being mounted to human subjects for the accurate measurement of clinical angles of skeletal orientation under varying ranges of dynamic movements. The IBME joint model including the dynamic calibration techniques provide a framework for the acquisition of limb motion transferred to a skeletal model which accommodates various limb lengths, sizes and shapes. The IBME joint model is stable and has been demonstrated to be portable to other sensors on the market to increase the repeatability of data acquired with surface mounted sensors

ACKNOWLEDGEMENTS

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