Assessing longitudinal change in coordination of the paretic upper limb using on-site 3-dimensional kinematic measurements


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Chapter 2

ABSTRACT

Background and Purpose. It is largely unknown how adaptive motor control of the paretic upper limb contributes to motor recovery after stroke. This paucity of knowledge emphasizes the need for longitudinal 3-dimensional (3D) kinematic studies with frequent measurements to establish changes in quality of motor control after stroke. A portable 3D kinematic setup would facilitate the frequent follow-up of patients with stroke. This case report shows how longitudinal 3D kinematic changes of the upper limb can be measured at a patient’s home using a portable 3D kinematic system in the first 6 months after stroke.

Case Description. The outcomes of the upper limb section of the Fugl-Meyer Motor Assessment (FMA), the Action Research Arm Test (ARAT), and 3D kinematic analyses were obtained from a 41-year-old man with a left hemispheric stroke. 3D kinematic data of the paretic upper limb were collected during a reach-to-grasp task using a portable motion tracker in 5 measurements during the first 6 months after stroke. Data from an individual who was healthy were used for comparison.

Outcomes. The FMA and ARAT scores showed most improvements in the first 5 weeks, accompanied by significant changes in 3D kinematic outcomes over time poststroke. Specifically, elbow extension increased, forward trunk motion decreased, peak hand speed increased, peak hand opening increased, and peak hand opening occurred sooner after peak hand speed.

Discussion. This case report illustrates the feasibility of frequently repeated, on-site 3D kinematic measurements of the paretic upper limb. Early after stroke, task performance was mainly driven by adaptive motor control, whereas adaptations were mostly reduced at 26 weeks after stroke. The presented approach allows the investigation of what is changing in quality of motor control and how these changes are related to the non-linear pattern of improvements in body functions and activities after stroke.
INTRODUCTION

Eighty percent of all patients with stroke show a reduced ability to use the paretic upper limb during activities of daily living.\(^1\) Longitudinal prospective cohort studies have shown that progress of time explains most improvements in upper limb functioning, which mainly occur in the first 2 months after stroke.\(^2,3\) Kwakkel and colleagues hypothesized that these time-dependent improvements reflect underlying neurological mechanisms such as salvation of penumbral tissue and alleviation of cerebral shock.\(^3\) Moreover, Krakauer showed that the brain can adapt to the neurological deficits by learning-dependent cortical reorganization, which may serve as a neuronal substrate for rehabilitation.\(^4\)

However, the meaning of these neurological processes for motor recovery after stroke is largely unknown, as besides restitution by true neurological recovery, substitution of function by compensatory movements may contribute to improvements in task performance.\(^5\) Substitution of function is indicated by, for instance, increased flexion of the trunk during forward reaching movements in patients with stroke.\(^6\) These trunk movements may reflect an adaptive control strategy to compensate for motor deficits in the paretic upper limb,\(^6\) such as exaggerated velocity-dependent tendon reflexes and increased stiffness (i.e., spasticity).\(^7\) The role of restitution and substitution of function is shown in Figure 2.1, representing a theoretical framework for underlying mechanisms of skill acquisition after stroke.

The appearance of compensatory movement patterns in patients with stroke reflects the redundancy in motor control. That is, muscles and joints can be used in numerous ways to accomplish a certain functional goal. Bernstein stated that, mechanically, the motor system has a redundant amount of degrees of freedom (DOFs).\(^8\) Coordination, in this view, is “the process of mastering the redundant DOFs, in other words, its conversion into a controllable system.”\(^8\), p127

Nijland and colleagues showed that the control over specific DOFs, such as shoulder abduction and finger extension, is essential for regaining arm and hand function in patients with stroke.\(^9\) In addition, patients may learn to optimize their motor control strategies. Based on several parameters, such as energy expenditure, speed, accuracy,\(^10\) and success,\(^11\) the motor system will gradually adapt its strategy to
master the DOFs until all relevant parameters are optimally adjusted. The ability to optimize motor control implies that motor compensations should not be considered pathological, but rather adaptive to existing motor impairments.

Figure 2.1 Mechanisms underlying motor recovery after stroke. Initially, stroke impairs the awareness of body movements (i.e., the ability to perceive the environment and to modulate motor control). In addition, it induces biomechanical changes in the upper limb. As a consequence, patients with stroke may show adaptations in motor control during the performance of functional tasks. Processes of spontaneous neurological recovery, such as salvation of penumbral tissue and alleviation of diaschisis, may lead to restitution of motor function and eventually to improvements in task performance. Motor learning may be considered as an alternative mechanism to regain functional ability and may induce substitution of motor function by compensatory movements.

To establish changes in quality of motor control and to understand how adaptive motor control contributes to improvements in functional ability, longitudinal studies need to include 3-dimensional (3D) kinematic or electromyographic analyses as part of the design, along with clinical measures of impairment. Furthermore, the early and
spontaneous functional improvements suggest that the dynamics between motor impairment and motor adaptations change mostly in the acute phase after stroke and stabilize after approximately 6 months after stroke. Therefore, measurements need to be repeated frequently, particularly at fixed time-points in the acute phase after stroke, to investigate the most dynamic period after stroke.

However, due to early discharge policies from hospitals to different settings, 3D kinematic measurements generally require patients to travel to specialized motion laboratories in university hospitals. Thus, 3D kinematic studies with frequent repeated measurements (i.e., intensive repeated measures designs) in the first 6 months after stroke are often perceived as burdensome for patients and their relatives. As a consequence, it is difficult to frequently follow patients with stroke in longitudinal studies at fixed time-points throughout their course of recovery. Follow-up of these individuals becomes even more difficult in multicenter studies because it requires equivalent 3D kinematic setups in each participating center. This may explain why previous longitudinal 3D kinematic studies into quality of motor control of the paretic upper limb were restricted to only 2 or 3 measurements at arbitrary time-points determined by time of admission or discharge.

The purpose of this case report was to investigate whether improvements in task performance could be explained by spontaneous restitution of motor function or learned adaptations in coordination. In addition, we examined whether frequently repeated 3D kinematic measurements can be conducted, using a portable measurement system, at fixed time-points irrespective of where patients reside during the first 6 months after stroke.

METHODS

Case description
Mr X was a 41-year-old man who was admitted to a hospital with a first-ever unilateral ischemic stroke in the left hemisphere, established by means of magnetic resonance imaging. The stroke was classified as a lacunar anterior cerebral infarction according to the Bamford Classification. The Edinburgh Handedness Inventory showed that the affected right arm was his dominant arm.
In the first week after his stroke, Mr X was able to make some synergistic movements with his paretic arm, whereas control over finger movements was affected. However, he showed some active finger extension at 5 days after stroke, providing a favourable prognosis for regaining arm-hand function after 6 months after stroke. For comparison, an healthy individual participated in one 3D kinematic measurement session and served as a control participant. This control participant was a right-handed, 43-year-old man.

This case report was part of the EXPLICIT-stroke programme and was approved by the local ethics committee. Both individuals gave written informed consent before participating.

Intervention and clinical measurements
Seven days after stroke, Mr X was discharged from the hospital and returned home. For the next 3 weeks, he continued visiting a rehabilitation center once a day where he received 1 hour of outpatient physical therapy in accordance with the Dutch guidelines for physical therapy. Initially, he trained only out-of-synergy movements. As he became more skilled in these movements after 1 week of therapy, the focus of the therapy gradually shifted to grasping movements and hand movements during simple activities of daily living, such as drinking and opening doors. After these 3 weeks of intensive therapy, Mr X received outpatient treatment with a physical therapist focusing on arm-hand movements for, on average, 2.5 hours per week. Treatments took place in a rehabilitation center without a motion analysis laboratory. Motor impairment and functional ability of the paretic arm were measured by means of the upper extremity part of the Fugl-Meyer Motor Assessment (FMA) and the Action Research Arm Test (ARAT) at weeks 1, 3, 5, 8, 12, and 26 after stroke.

3D kinematic measurement procedure
All 3D kinematic measurements were performed at Mr X’s home, which eliminated the burden of traveling to another medical facility with a motion laboratory. 3D kinematic data were not recorded in the first week after stroke due to Mr X’s lack of ability to perform reach-to-grasp movements. The 3D kinematic measurement of the control participant was conducted in the VU University Medical Center.
**Instrumentation**

Both individuals were seated at a portable table with a height of 76 cm. The 3D kinematic data (i.e., position and orientation) were collected with a portable electromagnetic motion tracker (Polhemus Liberty, Polhemus, Colchester, Vermont). The weight of the portable motion analysis system (including sensors and magnetic source) was 15 kg. All measurements were obtained relative to a global reference frame with its origin at the center of the magnetic source, the x-axis directed forward, the y-axis directed upward, and the z-axis directed rightward. The position of the magnetic source is depicted in Figure 2.2. The sampling frequency was 240 Hz. Using double-sided adhesive tape, the motion sensors were attached to the thorax, scapula, upper arm, forearm, hand, and the nails of the thumb and index finger.

In addition, an anatomical calibration procedure was executed before each measurement.

![Figure 2.2 Task execution. The person starts in the initial position (left panel). The person reaches for the block (small black square) at the block position (middle panel) and places the block at the end position (right panel). The magnetic source is represented by the large black square. The small rectangles on the patient (left panel) indicate the position of the sensors. The dashed line represents the maximum reaching distance of the non-paretic arm.](image)

In this procedure, the position of each of 13 anatomical landmarks relative to its associated sensor was digitized using a pointer device or stylus (ST8, Polhemus, Colchester, Vermont). In addition, the location of the glenohumeral joint was calculated using linear regression from the scapular landmarks.\(^1\) The anatomical landmarks are listed in Table 2.A2 in the Appendix.

All 3D kinematic data were collected within a range of 60 cm from the source. The Appendix shows that: (1) within this range, measurement environment does not affect the accuracy of the 3D kinematic data, and (2) minimal variations in the localization of the anatomical landmarks hardly affect the reliability of the 3D kinematic outcomes.
Chapter 2

Paradigm
The functionality of upper-limb movements largely depends on the extent to which objects can be grasped. Therefore, 3D kinematic data were collected during a reach-to-grasp movement using a cubic block (150 g) with an edge length of 5 cm. Both individuals performed the task with the right arm, which was the paretic arm for Mr X. In the initial position, the back was against the backrest of the chair and both feet were on the floor. The distance between the chair and the table was kept constant over all measurements. The initial hand position was on the edge of the table in front of the shoulder, keeping the tips of the thumb and index finger together. Both individuals were instructed to grasp a block at comfortable speed after the instructor gave a verbal ‘GO’ signal. The block position was in front of the hand in its initial position on a line parallel to the front edge of the table that could be reached with the fully extended less-affected arm while keeping the back against the chair. Having picked up the block, the individual had to transport it to the end position, which was located at the left of the initial block position at a distance equal to the distance between the initial hand position and the block position. This task is also depicted in Figure 2.2. After the ‘GO’ signal, both individuals were allowed to move their trunk from the back of the chair in case this position was more comfortable. Whenever the block was dropped after it was picked up, the recording was deleted and an extra trial was added. The task was repeated 7 times during each 3D kinematic measurement.

All analyses were performed on the reaching phase, that is, from movement onset until the block lost contact with the table. The transport of the block to the left block position was excluded from the analyses, but was required for the determination of the end of the movement. Beginning and end of movement were defined, respectively, as the moment of movement onset and block lift and were determined from the linear velocity of the wrist sensor. Linear velocity of the wrist sensor was calculated using the first derivative of the sensor displacement relative to the magnetic source. This signal was filtered with a low-pass second-order Butterworth filter (recursive filter, cut-off frequency = 20 Hz). Movement onset was defined as the moment at which the velocity exceeded 5% of the maximum velocity during the reaching phase. Moment of block lift was defined as the moment at which the velocity exceeded 5% of the maximum velocity during the block transport phase.
Data analysis
Trunk and hand displacements were based on displacements of the upper side of the sternum (i.e., the incissura jugularis) and the base of the third metacarpal bone, respectively. Elbow rotations were calculated according to the recommendations of the International Society of Biomechanics. Specifically, elbow flexion and elbow extension were defined as positive and negative rotation, respectively, of the forearm about the transversal axis of the upper arm (i.e., the axis through the medial and the lateral epicondyle of the upper arm). The data analysis was executed using an adapted version of BodyMech 3.06.01 (VU University Medical Center, Amsterdam, the Netherlands).

Outcome parameters
The 3D kinematic parameters were selected based on Jeannerod’s model to describe reach-to-grasp movements. This model assumes 2 components that are coordinated by 2 interdependent visuomotor control channels: the transport component and the grasp component. The transport component brings the hand in the vicinity of the object, whereas the grasp component preshapes the hand and fingers and finally encloses the object. Both components occur simultaneously and are temporally coupled.

Patients with stroke tend to perform reach-to-grasp movements with reduced elbow extension. Michaelsen and colleagues hypothesized that patients use increased forward trunk motion to compensate for reduced elbow extension. Elbow flexion and trunk forward displacement, therefore, are important to describe the dependency on synergistic upper limb movements and compensatory trunk movements.

However, compensatory trunk movements require more body mass displacement and, therefore, are less energy-efficient. Maximum hand speed was used to investigate whether these less energy-efficient strategies are accompanied by slower hand transportation.

Furthermore, impaired control over the finger extensors may lead to an insufficient opening between the fingers and the thumb, which impedes preshaping of the hand around an object. Peak hand opening, therefore, is a measure that provides insight into the function of the finger extensors during the reach-to-grasp movement.
Although the transport and grasp component occur simultaneously in healthy individuals, there are indications that patients with stroke cannot control multiple processes in parallel, and prefer a serial processing of movements.\textsuperscript{24} The relative timing of the maximum hand speed and maximum hand opening was used as a measure of temporal coupling between the hand transport component and the grasp component. Specifically, the 3D kinematic outcome parameters were defined as follows.

Global variable

- Movement duration: time between movement onset and block lift.

Transport component

- Forward trunk displacement: displacement of the incissura jugularis in a forward direction.
- Elbow angle: elbow flexion angle at the moment of block lift. Full elbow extension is represented by 0 degrees.
- Peak hand speed: maximum linear velocity of the base of the third metacarpal bone.

Grasp component

- Peak hand opening: maximum distance between the tip of the thumb and the tip of the index finger during the movement.
- Timing of hand opening (THO): moment of peak hand opening (tPHO) relative to the moment of peak hand speed (tPHS), calculated by the following equation:

$$THO = \frac{t_{PHO}}{t_{PHS}} \times 100\%$$ \hspace{1cm} 2.1

All outcome parameters were calculated using custom-written Matlab software (version R2006a, MathWorks Inc, Natick, Massachusetts).
Statistics
Visual inspection was used to detect differences between Mr X and the control participant and patterns in the changes of 3D kinematic outcomes over time poststroke. Furthermore, single case statistics were applied to determine whether the mean 3D kinematic outcomes over the 7 repeated reach-to-grasp movements within each measurement changed significantly over time poststroke and between consecutive measurements. Specifically, the non-parametric Kruskal-Wallis analysis of variance (SPSS version 15.0, IBM Corporation, Armonk, New York) was used for each 3D kinematic outcome. Significance level was set at $P \leq 0.05$. In case of significance, the Mann-Whitney U test (2-tailed) was used to check whether outcomes obtained from consecutive assessments differed significantly from each other. The Bonferroni adjustment for multiple testing was applied. This adjustment implies that the level of significance was divided by the number of Mann-Whitney U tests per outcome variable (i.e., 4), which resulted in a corrected level of significance of $P \leq 0.05/4 = 0.0125$.

RESULTS
Motor function and functional ability of the paretic upper arm were assessed with, respectively, the upper extremity part of the FMA and the ARAT. Figure 2.3 shows that Mr X’s scored 23 points on the FMA in the first week after stroke indicating that he was able to make synergistic movements with his paretic arm. However, his score of 3 points on the ARAT indicated that the paretic upper limb could not be used for functional purposes.

Both the FMA and ARAT scores showed most improvements in the first 5 weeks after stroke whereas the maximum score was reached around 8 weeks after stroke. The average time required to conduct a full 3D kinematic measurement was 1.5 hours (range = 1.3 – 1.8) (including the installation of the setup and anatomical calibration).
**Figure 2.3** Improvement in motor function, as indicated by scores on the upper extremity part of the Fugl-Meyer Motor Assessment, and functional recovery, as indicated by scores on the Action Research Arm Test. The absolute scores for both clinical outcome measures are presented in the figure.

Figure 2.4 illustrates that each 3D kinematic outcome changed significantly as a function of time poststroke (P < 0.001). Moreover, significant changes were detected between some consecutive measurements. Some 3D kinematic outcomes showed further changes beyond 8 weeks after stroke, despite the fact that the maximum score on the FMA and ARAT was achieved at this time point.

Movement duration as a global parameter decreased over time poststroke. At 26 weeks after stroke, movement duration was almost equivalent for Mr X and the control participant. Changes in the control of the reach-to-grasp movement were investigated using 3D kinematic outcomes.
Kinematics of reaching after stroke

Figure 2.4 Kinematic outcomes for the patient with stroke and the control participant. Each circle represents the outcome for one trial. Asterisks indicate a significant change with respect to the previous measurement. The outcomes of the control participant for each trial are marked with an ‘x’ at the right of each graph.
Transport component

As a function of time poststroke, forward trunk displacement and elbow angle changed toward the values obtained from the control participant. However, Figure 2.4 shows that, with respect to elbow angle, the difference between Mr X and the control participant remained relatively large.

Figure 2.5 displays the elbow angle and forward trunk displacement during the reach-to-grasp movement at weeks 3, 5, and 26 after stroke onset. The range of motion in the elbow increased over time poststroke, and the amount of trunk movement diminished. Visual inspection also showed that the trajectories became smoother and more consistent.

The mean peak hand speed increased from 0.57 m/s in week 3 to 0.82 m/s in week 26 after stroke, indicating that the transport of the hand occurs faster as motor function improves. Moreover, visual inspection showed that mean peak hand speed in week 3 was smaller than mean peak hand speed in the control participant, whereas in week 26 after stroke the opposite was the case.

Figure 2.5 Graphic representation of the increase in elbow range of motion and the simultaneous decrease of trunk forward motion as a function of time poststroke.
Grasp component
The mean peak hand opening increased from 6.4 cm in week 3 to 11.5 cm in week 12 and subsequently decreased to 10.2 cm in week 26. Figure 2.4 shows that, compared with the control participant, Mr X’s peak hand opening was smaller in week 3 after stroke, whereas his peak hand opening was larger in week 26 after stroke. In addition, the timing of peak hand opening relative to peak hand speed is represented by timing of hand opening, which decreased mostly early after stroke.

DISCUSSION
To our knowledge, the present case report is the first to conduct intensively repeated 3D kinematic measurements of the paretic upper limb outside the laboratory during the first 6 months after stroke. Detailed 3D kinematic data were collected within, on average, 1.5 hours per assessment. Although this time is longer than the time required to administer most clinical tests such as the upper extremity part of the FMA and the ARAT, the portable setup facilitates frequent 3D kinematic assessments of reach-to-grasp movements in patients with stroke over the course of recovery, which is considered as a major advantage.

Clinical assessments indicated that motor function and functional ability improved mainly in the first 5 weeks, with an apparent plateau of recovery at the maximum score around 8 weeks after stroke. By contrast, most 3D kinematic outcomes indicated ongoing recovery up to attainment of performance levels relatively comparable to those of the control participant at 6 months after stroke. This finding suggests that standard clinical assessments used in clinical trials may not be sufficiently sensitive to capture further improvement of motor function due to a ceiling effect. 3D kinematic outcomes, as presented in this case report, therefore, may be a valuable addition to these trials because they do not appear to exhibit this ceiling effect and could detect actual changes in motor function beyond the apparent plateau. The apparent absence of a ceiling effect would be particularly valuable in patients who have a favourable prognosis for regaining dexterity based on active finger extension in the first week after stroke, as was the case with the patient described in this case report. Furthermore, clinical assessments do not specify how quality of motor
control changes over time poststroke and how it contributes to improvements in functional ability.

The 3D kinematic measurements revealed that in week 3 after stroke, the trunk made a large contribution to the transport of the hand. However, as time progressed, trunk movements diminished and the elbow was gradually more extended during the reach-to-grasp task. There are several possible causes for this change in motor control. First, it has been shown that afferent joint feedback is most accurate when joints are around the mid-position. Since patients with stroke may need more accurate feedback to steer their movements, they might prefer to keep the elbow more flexed. Second, increased elbow extension might be the result of improved strength in the elbow extensors. As a consequence, forward trunk movements may become unnecessary. Third, the initially observed increased contribution of the trunk combined with a reduced contribution of the elbow suggests that this patient froze distal degrees of freedom. This phenomenon is typical of inexperienced learners and enables them to control fewer (proximal) degrees of freedom, which facilitates motor control. As motor control improves during motor learning, freeing of distal degrees of freedom, such as in the elbow, occurs, leading to more efficient motor control strategies.

With respect to the grasp component, the maximum hand opening during the task increased significantly during the first 12 weeks after stroke. The parallel increase in the score on the upper extremity part of the FMA suggests that this increase is caused by a return of selective control over the fingers. Interestingly, maximum hand opening decreased between 12 and 26 weeks after stroke, which suggests that this patient with stroke became more skilled in reach-to-grasp movements and that a smaller hand opening was sufficient for a successful grasp. Larger hand openings to ensure a successful enclosure of the fingers are also observed in healthy subjects who are instructed to perform reach-to-grasp movements faster than their preferred speed.

Furthermore, the peak hand opening occurred sooner after peak hand speed as time poststroke increased. This finding suggests that the grasp and the transport component became more parallel, rather than serial controlled, which is indicative of improved quality of motor control.
Based on the findings of this case report, we concluded that these frequently repeated, on-site 3D kinematic measurements increase our insight into what is changing in quality of motor control during early improvements in body functions and activities after stroke.\textsuperscript{3,5} This insight provides the opportunity to determine whether patients with stroke show true motor recovery or learn to compensate for impairments, even though based on clinical measures, full recovery might be assessed, as in this case report. In particular, the present setup may serve as a measurement tool to assess the quality of motor control in controlled trials to determine the effectiveness of interventions that are aimed at normalizing intralimb movement coordination, such as neurodevelopmental treatment and shoulder-elbow robotics.\textsuperscript{28,29} Furthermore, these measurements avoid the disadvantages of fixed setups and fit well with recent rehabilitation policies promoting early discharge to the community.\textsuperscript{30}

This case report was part of a pilot study for the EXPLICIT-stroke programme. EXPLICIT-stroke is a multicenter translational research programme that aims to investigate the effects of early applied intensive intervention for regaining dexterity after stroke.\textsuperscript{15} Frequent 3D kinematic measurements are used to investigate whether improvements in task performance are the result of spontaneous restitution of motor function or substitution of function by compensation.\textsuperscript{15} Randomized controlled trials conducted in the first few weeks after stroke, with repeated measurements over time using multiple measurement techniques to assess various body functions and activities, should provide improved insights into the underlying time-dependent and learning-dependent mechanisms of motor recovery after stroke.\textsuperscript{2}
APPENDIX

Background
The quality of three-dimensional kinematic data collected by electromagnetic motion trackers is affected by magnetic distortion induced by metal objects in the environment, such as metal frames in furniture and concrete walls. Thus, on-site 3D kinematic measurements are potentially subject to magnetic distortion that may vary across measurement environments. In addition, the reliability of 3D kinematic outcomes depends on the accuracy of the anatomical calibration. Therefore, it is desirable to reduce variations in the localization of the anatomical landmarks between and within assessors as much as possible within longitudinal studies, especially when they concern multicenter trials.

The data within this Appendix specify: (1) the accuracy of position and rotation data in different measurement environments and (2) the effect of variations in the anatomical calibration on the reliability of the 3D kinematic outcomes. Both experiments were conducted using the Polhemus Liberty motion tracker (Polhemus, Colchester, Vermont).

Accuracy of the raw data in different measurement environments
This experiment was conducted in 3 different environments: (1) a motion laboratory in a medical center, (2) a treatment room in a rehabilitation center, and (3) a home situation. In each environment, position and orientation were measured at 35 locations by means of the frame depicted in Figure 2.A1. The error in the position data was expressed as the vector distance between the known position and the measured position. The error in orientation data was expressed as the magnitude of the rotation about the screw axis from the known orientation to the measured orientation.\textsuperscript{31,32}
Figure 2.A1 The frame, made of Plexiglas, that was used to measure position and orientation at specific locations in the measurement environment. Seven motion sensors were fixed onto the frame, and their orientation was exactly equal to the global reference frame with its origin in the center of the magnetic source (gray).

Figure 2.A2 shows the error in the position and orientation data as a function of the distance to the magnetic source in each measurement environment. Visual inspection reveals that the error in measured position is equivalent throughout the required measurement range of 60 cm to the source, whereas errors increase rapidly as the sensors are located further away from the source. Likewise, the error in orientation data increases rapidly when the distance from the source is larger than 60 cm. However, errors in orientation data are also large when the distance to the source is smaller than 20 cm from the center of the source. Table 2.A1 presents the mean errors in position and orientation within their most accurate ranges.

Because upper limb movements during the presented reach-to-grasp task are recorded within 60 cm from the source, we concluded that the presented setup allows collection of accurate data in different environments.
Figure 2.A2 Left panel: vector distance from the measured to the actual position as a function of distance to the source. Right panel: rotation about the screw axis from the measured to the actual orientation as a function of distance to the source. Large dots represent the motion laboratory, triangles represent the rehabilitation center, small dots represent the home situation.

Table 2.A1 Mean [range] error in position and orientation data in different environments

<table>
<thead>
<tr>
<th>Environment</th>
<th>Error in position data (cm)</th>
<th>Error in orientation data (°)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Motion laboratory</td>
<td>0.65 [0.21 – 1.46]</td>
<td>2.0 [0.0 – 3.1]</td>
</tr>
<tr>
<td>Treatment room</td>
<td>0.76 [0.34 – 1.43]</td>
<td>1.8 [0.4 – 2.3]</td>
</tr>
<tr>
<td>Home situation</td>
<td>0.83 [0.31 – 1.85]</td>
<td>2.7 [1.1 – 4.0]</td>
</tr>
</tbody>
</table>

1. Concerns data collected within 60 cm from the centre of the magnetic source.
2. Concerns data collected between 20 and 60 cm from the centre of the magnetic source.

Effect of variations in anatomical calibration on outcome parameters

Ten individuals (5 male, 5 female) performed 7 repetitions of the reach-to-grasp movement described in the present case report while upper limb 3D kinematics were recorded at 240 Hz. The task was performed with the right arm by all individuals. Prior to each measurement, 2 assessors performed an anatomical calibration in which the positions of 13 anatomical landmarks were digitized. Table 2.A2 shows the list of anatomical landmarks. One of the assessors performed the anatomical calibration twice. The purpose of this experiment was to compare the 3D kinematic outcomes that were obtained from the same motion recordings but from different anatomical calibrations. The following 3D kinematic outcomes were calculated: forward trunk displacement, elbow angle, peak hand speed, peak hand opening, and timing of peak hand opening. The definitions of these outcomes are presented in the case report. For every parameter, the mean over the 7 repeated reach-to-grasp movements was used for further analysis.
Table 2.A2 List of Anatomical Landmarks

<table>
<thead>
<tr>
<th>Anatomical landmarks</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>Thorax</td>
<td></td>
</tr>
<tr>
<td>IJ: incissura jugularis</td>
<td>Deepest point (supra sternal notch)</td>
</tr>
<tr>
<td>Scapula</td>
<td></td>
</tr>
<tr>
<td>TS: trigonum spinae</td>
<td>Midpoint of the triangular surface on the medial border of the scapula in line with the scapular spine</td>
</tr>
<tr>
<td>AI: angulus inferior</td>
<td>Most caudal point of the scapula</td>
</tr>
<tr>
<td>AA: angulus acromialis</td>
<td>Most laterodorsal point of the scapula</td>
</tr>
<tr>
<td>PC: processus coracoideus</td>
<td>Most ventral point of the scapula</td>
</tr>
<tr>
<td>AC: acromio-clavicular joint</td>
<td>Most dorsal point of the acromio-clavicular joint</td>
</tr>
<tr>
<td>Humerus</td>
<td></td>
</tr>
<tr>
<td>GH: glenohumeral rotation center *</td>
<td>Rotation center of the glenohumeral joint</td>
</tr>
<tr>
<td>EL: lateral epicondyle</td>
<td>Most caudal point on the EL</td>
</tr>
<tr>
<td>EM: medial epicondyle</td>
<td>Most caudal point on the EM</td>
</tr>
<tr>
<td>Forearm</td>
<td></td>
</tr>
<tr>
<td>US: ulnar styloid</td>
<td>Most caudal and medial point on the US</td>
</tr>
<tr>
<td>RS: radial styloid</td>
<td>Most caudal and lateral point on the RS</td>
</tr>
<tr>
<td>Hand</td>
<td></td>
</tr>
<tr>
<td>MC3: processus styloideus os metacarpal 3</td>
<td>Most dorsal point on dorsal side of the hand</td>
</tr>
<tr>
<td>Thumb</td>
<td></td>
</tr>
<tr>
<td>TT: tip of the thumb</td>
<td>Most distal point of the thumb</td>
</tr>
<tr>
<td>Index finger</td>
<td></td>
</tr>
<tr>
<td>TI: tip of the index finger</td>
<td>Most distal point of the index finger</td>
</tr>
</tbody>
</table>

* determined by means of linear regression from the scapular landmarks

For each mean outcome parameter, the mean difference, the 95 % confidence interval, and the 95 % limits of agreement were calculated between and within assessors. In addition, interrater reliability was calculated using intraclass correlation coefficients with a 2-way random effects model with an absolute agreement definition. Similarly, the intrarater reliability was determined by applying a 2-way mixed effects model for absolute agreement. Statistics for this experiment were calculated using SPSS version 15.0 (IBM Corporation, Armonk, New York).

Table 2.A3 shows the mean difference, the 95 % confidence interval, the 95 % limits of agreement, and the intraclass correlation coefficients for the different outcome parameters.
Table 2.A3 Inter- and intrarater reliability

<table>
<thead>
<tr>
<th></th>
<th>Inter</th>
<th>95 % CI</th>
<th>95 % LOA</th>
<th>Intrarater</th>
<th>95 % CI</th>
<th>95 % LOA</th>
<th>ICC</th>
</tr>
</thead>
<tbody>
<tr>
<td>FTD (cm)</td>
<td>-0.14</td>
<td>-0.52 – 0.23</td>
<td>-1.17 – 0.90</td>
<td>0.86</td>
<td>-0.09</td>
<td>-0.49 – 0.31</td>
<td>1.19</td>
</tr>
<tr>
<td>EA (°)</td>
<td>-0.61</td>
<td>-4.84 – 3.62</td>
<td>-12.20 – 11.00</td>
<td>0.84</td>
<td>-0.51</td>
<td>-3.42 – 2.40</td>
<td>8.48</td>
</tr>
<tr>
<td>PHS (cm/s)</td>
<td>1.36</td>
<td>0.53 – 2.18</td>
<td>-0.92 – 3.64</td>
<td>0.99</td>
<td>0.04</td>
<td>-0.17 – 0.26</td>
<td>0.57</td>
</tr>
<tr>
<td>PHO (cm)</td>
<td>0.05</td>
<td>-0.21 – 0.31</td>
<td>-0.65 – 0.76</td>
<td>0.95</td>
<td>-0.05</td>
<td>-0.30 – 0.20</td>
<td>0.73</td>
</tr>
<tr>
<td>THO (%)</td>
<td>0.66</td>
<td>-2.86 – 4.18</td>
<td>-8.98 – 10.30</td>
<td>0.97</td>
<td>-1.07</td>
<td>-3.54 – 1.41</td>
<td>7.85</td>
</tr>
</tbody>
</table>

MD: Mean Difference; 95 % CI: 95 % Confidence Interval; 95 % LOA: 95 % Limits of Agreement of the Mean Difference; ICC: Intraclass Correlation Coefficient
FTD: Forward Trunk Displacement; EA: Elbow Angle; PHS: Peak Hand Speed; PHO: Peak Hand Opening; THO: Timing Hand Opening

The mean differences were close to zero, which indicated minimal systematic error between and within assessors. This finding was confirmed by the 95 % confidence interval, including zero, for all outcome parameters except for the mean difference in peak hand speed between assessors. In addition, interrater and intrarater reliability was high to excellent, as indicated by the intraclass correlation coefficients being larger than .8 for all outcome parameters. Intrarater reliability was generally larger than interrater reliability, which was confirmed by the smaller limits of agreement for 3D kinematic outcomes obtained from one assessor compared with those obtained by 2 assessors. These data confirm that the influence of variations in the localization of anatomical landmarks on the reliability of 3D kinematic outcomes is minimal.
REFERENCES


