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A portable magnetic position and orientation tracker

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Abstract

This paper presents the design and testing of a portable magnetic system for human motion tracking. Three essential components comprise this system (1) 3D source, consisting of three orthogonal coils, placed on the body; (2) a compatible 3D sensor, which is fixed at a remote body segment and measures the fields generated by the source; (3) a processor whose function is to relate the signals from source and sensor. Given the signals from the source and sensor, the position and orientation of the sensor in 6 degrees of freedom (DOF) with respect to the position of the transmitter can be estimated. Coil parameters, such as the radius are optimized for tracking the distance between the lower back and the shoulder of a person. The electronics are designed to run on battery supply, making it suitable for body mounting and ambulatory measurements. The system is tested with functional body movements such as flexion of the back, arm movement and walking. The accuracy of measurements is approximately 8 mm in position and 5° in orientation with 6 DOF movements. Results are less accurate during relatively high velocities of the source or sensor due to under sampling. The magnetic tracker will be used as an aiding system for fusion with inertial sensors, therefore, the update rate requirements can be relatively low.

Keywords: Magnetic tracker; Human motion; Ambulant

1. Introduction

In many biomechanical and virtual reality applications, position measurements on the body are important. For motion analysis laboratories, many systems are available based on different physical principles; for example, optical (e.g. Vicon; Oxford Metrics, Optotrak; Northern Digital), magnetic (e.g. Flock of Birds; Ascension, Star Track; Polhemus) or ultrasonic (e.g. IS300, InterSense). Under ambulatory conditions, possibilities for position measurements are limited. Miniature inertial sensors have been proposed for measurements outside the laboratory [16]. Positions and relative distances on the body can be estimated by using anatomic knowledge of segment lengths and joint characteristics in combination with inertial sensor based segment orientation estimates [1,23,15]. However, this approach is not satisfactory in cases with complex joints and non-rigid body segments such as the back and shoulder. Distances between body segments can principally not be assessed by numerical integration of the measured accelerations due to the unknown starting position. Only short-term estimates of position changes within seconds can be estimated because of the inherent drift associated with double integration of accelerations [7].

To estimate on-body positions accurately, inertial measurements need to be combined with an aiding method. In traditional navigation applications, the fusion of inertial sensors with aiding sources such as GPS or Doppler radar is well established. Foxlin et al. [6] fused acoustic time of flight measurements with miniature accelerometers and gyroscopes for 6 degrees of freedom (DOF) motion tracking. Emura and Tachi [5] combined a magnetic position and orientation tracking system with rate gyroscopes to improve the data rate and latency of a magnetic system. In these studies, a fixed lab system was combined with inertial measurements. In this present study, we will develop a portable magnetic tracking system, to be used as an aiding system. The magnetic-transducing technique overcomes occlusion problems associated with optical and acoustic tracking technologies [9].

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Magnetic trackers use an electromagnetic field generated at some point in space and detected at a remote segment [10,11]. Three essential components comprise these systems [19]:

- a 3D source, which generates a magnetic field;
- a compatible 3D sensor, which is fixed at a remote body segment and measures the fields generated by the source;
- a processor whose function is to relate the signals from source and sensor.

Given the signals from the source and sensor, the position and orientation of the sensor in 6 DOF with respect to the position of the transmitter can be estimated.

Commercially available magnetic trackers such as Fastrak (Polhemus) and Flock of Birds (Ascension Technology) are provided with so-called long or extended range sources offering a tracking range of several meters [3]. However, the source, consisting of large (30–45 cm diameter) 3D coils, is fixed in one place and therefore limiting the measurement volume. For biomechanical analysis in ambulatory settings, we are interested in relative distances between body segments. These relative distances are generally smaller than the distances from a fixed source to a moving sensor in a lab environment. Consequently, the required magnetic fields to bridge distances on the body are smaller. Therefore, the source and power supply can be scaled down to be attached to the body, making the system portable.

This paper focuses on the design of a portable magnetic system. Major requirements for such a system are small weight and size, and no impediment of functional mobility. It will be used as an aiding system for fusion with inertial measurements, therefore, the update rate requirements can be relatively low. The position estimates using miniature inertial sensors have an accuracy of a few percent after 1 s [8], therefore an update rate of 1–2 Hz of the magnetic system is sufficient. The operating time should be at least half an hour on a set of batteries, but preferably longer.

The calculations to resolve the 6 DOF are based on a magnetic dipole approximation of the source [11,19]. In the design of the coils we try to imitate the ideal, infinitely small dipole. Such imitation, however, is never perfect and causes inevitable errors, which dramatically increase at distances comparable with the coil dimensions [18]. In this paper, coil parameters, such as the radius are optimized for tracking the distance between the lower back and the shoulder of a person. With these results, the accuracy of the implemented magnetic distance and orientation estimates are evaluated by an optical reference system.

2. Design of the system

Fig. 1 shows a scheme of the implemented ambulatory magnetic system. The 3D source is constructed as a three orthogonally sided pyramid (base diameter = 21 cm, height = 11 cm, weight = 450 g). It is mounted on the back of the body and sensors are placed at remote body segments. The transmitter driver provides controlled pulsed dc current to three coils having orthogonal axes. The three-axis magnetic sensor measures the strengths of the magnetic pulses that are functions of the distance to the transmitter. The equations presented by Kuipers [12] are used to calculate the 6 DOF. The three position parameters are expressed in spherical coordinates, where R is the distance between source and sensor and α and β are the tracking angles between the source and sensor frames. The orientation between source and sensor is expressed by rotation matrix $\mathbf{\Psi}$. Fig. 2 shows the timing relationship of two identical cycles between the three orthogonal sources and sensors. During the period B_1 to B_2 the X-source is activated, from B_2 to B_3 the Y-source and from B_3 to B_1 the Z-source. Between the magnetic pulses, only the earth magnetic field is measured which can be subtracted from the measured pulses yielding the field $\mathbf{B}$ as emitted by the dipole-source. At the end of the cycle, nine values represent the relation

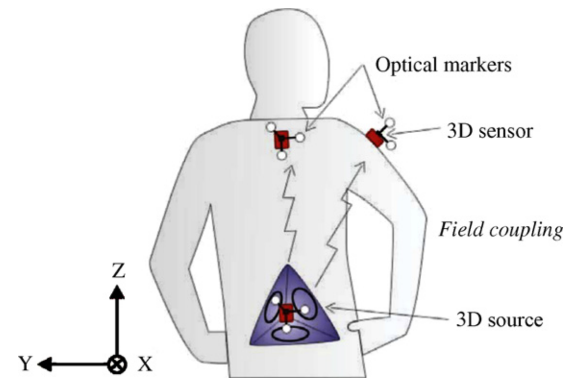

Fig. 1. Body-mounted magnetic system for measurement of relative distances and orientations on the body, consisting of a three-axis magnetic dipole-source worn by the subject and three-axis magnetic and inertial sensors on remote body segments.

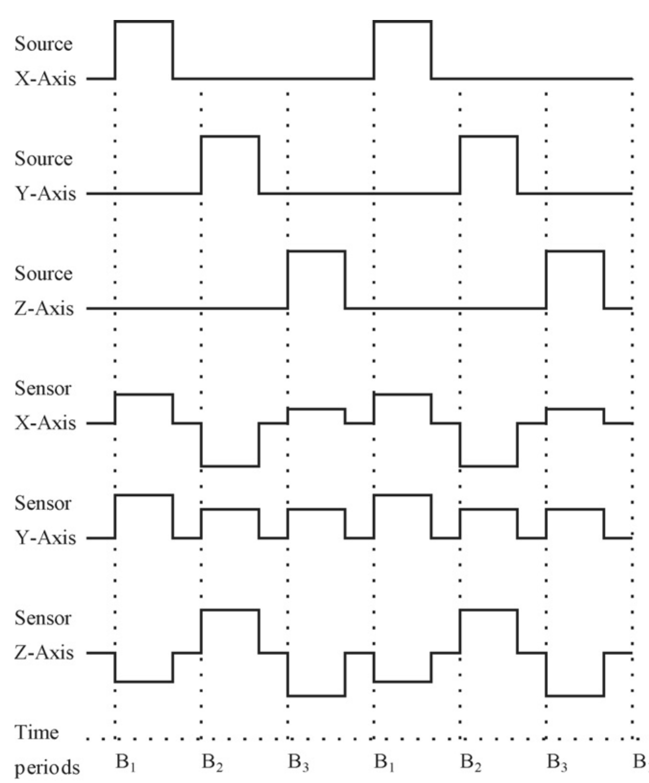

Fig. 2. Timing diagram showing the relationship between the transmitted and received signals.

between source and sensor; three sensor values for each time one of the coils is actuated. The entire cycle of pulsing the sources X, Y and Z repeats as long as measurements are required.

The system uses magnetic pulses because time-varying (ac) fields cause an induced electromotive force (emf) when magnetic flux flows in nearby conducting and ferrous materials. The induced emf will produce local currents in the materials normal to the magnetic flux in accordance with Faraday's law. These so-called eddy currents generate secondary magnetic field that will influence the magnetic distance measurement. When using pulses, eddy currents die out at an exponential rate after the pulse reaches its steady state value. Sampling the transmitted signals farther from the leading edge will result in a sensed signal containing less eddy current components. The relative sensitivity to ferromagnetic materials depends on the size and type of the metal [17,13].

Because the source and sensors are placed on the body, the absolute position and orientation of the magnetic system are not known. To determine the orientation $\mathbf{\Phi}$ of the source with respect to the global reference system, an inertial and magnetic sensor was attached to the source. Accelerometers provide a means to estimate inclination [14]. The magnetometers give information about the heading direction, when not measuring the magnetic pulses. Changes in angles can be determined by integration of angular velocity, provided by gyroscopes. In between the magnetic actuation, the orientation of the sensor module was calculated using the Kalman filter fusion algorithm as presented by Ref. [20].

Assuming the maximum distance to be covered on the body, the noise levels of a typical magnetic sensor and a sufficient signal-to-noise ratio (SNR), the minimal strength of the magnetic field can be calculated. The calculations and design of a coil to generate this field are presented in the next session.

2.1. Magnetic dipole

Fig. 3 shows a circular loop of wire with radius b that carries current I. The magnetic potential $\mathbf{A}$ at a distance R_1 can be found by applying the Biot–Savart law [2]:

$$\mathbf{A} = \mathbf{a}_\phi \frac{\mu_0 I b}{2\pi} \int_{-\pi/2}^{\pi/2} \frac{b \sin\phi'}{R_1} \mathrm{d}\phi' \tag{1}$$

with μ_0 being the magnetic permeability of vacuum ($4\pi \times 10^{-7}\,\mathrm{T\,m^2/A}$). The magnetic flux density is $\mathbf{B} = \nabla \times \mathbf{A}$. To calculate the magnetic field when $R \gg b$, a coil can be considered as a magnetic dipole, with the magnetic induction $\mathbf{B}$ being expressed in spherical coordinates:

$$\mathbf{B}_{\mathrm{dipole}} = \frac{\mu_0 M}{4\pi R^3}(\mathbf{a}_R 2\cos\varphi + \boldsymbol{a}_\varphi \sin\varphi) \tag{2}$$

where M is the magnetic dipole:

$$M = NI\pi b^2 \tag{3}$$

The axis of the coil is aligned with the line $\varphi = 0$, $\mathbf{a}_R$ and $\mathbf{a}_\varphi$ are radial and tangential unit vectors. N is the number of turns of wire in the coil. Fig. 4 shows the relative error between the magnetic field as emitted by a coil and equivalent dipole, as a function of the relative distance R with respect to radius b. The error $\varepsilon_{\mathrm{approx}}$ is given by:

$$\varepsilon_{\mathrm{approx}} = \frac{|\mathbf{B}_{\mathrm{coil}} - \mathbf{B}_{\mathrm{dipole}}|}{\mathbf{B}_{\mathrm{coil}}} \times 100\% \tag{4}$$

In the dipole approximation, three parameters need to be optimized: the coil radius b, the number of windings N and the current through the coil I. Although a coil with a large area has a higher magnetic moment M according to Eq. (3), the accuracy in dipole approximation will decrease. Increasing the number of windings also increases M, however, it will result in a higher self-inductance and resistance, which is undesirable, especially in case of battery power supply. A higher resistance requires a higher voltage to drive the same amount of current through the coil. An increased self-inductance results in a longer rise time of the applied pulse. Finally, the current through the coil is partly limited by the internal resistance of the battery and partly by its capacity, the additional weight and maximum measurement

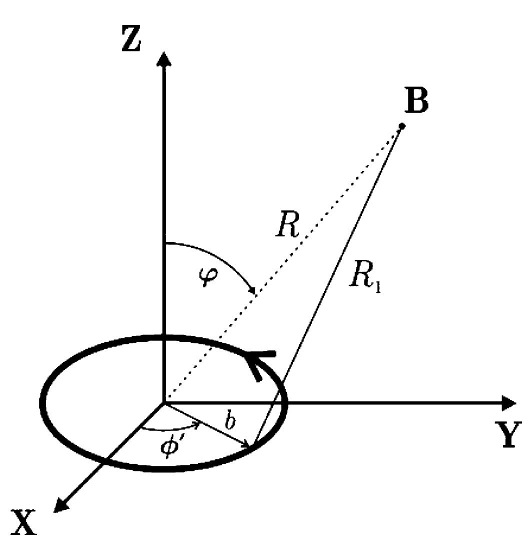

Fig. 3. A circular loop of wire, with radius b, carrying current I.

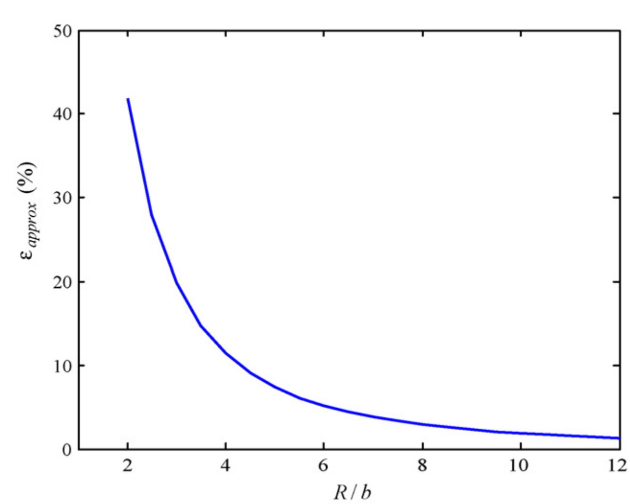

Fig. 4. Relative error $\varepsilon_{\mathrm{approx}}$ of a coil compared to a dipole as a function of distance from the coil along the Z-axis.

time. From Fig. 4, we find that the systematic dipole approximation error decreases as the distance from the coil increases, however at larger distances, stochastic errors will determine the accuracy of the 6 DOF measurement.

The required magnetic dipole strength M assuming a SNR n can be obtained using:

$$M = \frac{4\pi R_{\max}^3 n \mathrm{B}_n}{\mu_0} \tag{5}$$

The maximum distance $R_{\max}$ to be covered by the magnetic field was based on the distance between the source as placed on the lower back and the shoulder and was assumed not to exceed 70 cm. The noise level B_n of the used magnetoresistive sensor, with the necessary electronic flipping circuit [21] is about 0.5×10^{-7} T [22]. With a SNR n of 4, the magnetic dipole becomes 0.69 A/m. The necessary accuracy in the dipole approximation was set at 5% at a distance half of the specified distance between the lower back and shoulder (=35 cm). In Fig. 4, we can find the corresponding coil radius b, which is 5.5 cm. The number of windings N and current I followed from the necessary field strength and were 50 and 1.5 A, respectively. The time constant (L/R) of the coil was 0.23 ms.

3. Experimental methods

An electrical circuit was designed to drive the coils by means of four AA (LR6—2400 mAh) batteries. The pulse duration, duty cycle and driving current could be controlled by means of a Bluetooth interface. The magnetometers in a MTx (Xsens Technologies) sensor module were used to measure the strength of the pulses and the earth magnetic field in 3D. The sample frequency of the sensors was 120 Hz with 16 bits resolution. The magnetometers in the sensor module are 'flipped' every few samples to reduce both sensor offset and temperature effects [21]. A Vicon 470 system (Oxford Metrix) consisting of six cameras operating at 120 Hz was used as a reference. Three optical markers with a diameter of 25 mm were attached to the sensor module in an orthogonal arrangement to validate the sensor's position and orientation with respect to the position of the coils as can be seen in Fig. 1. The reference system was assumed to have an accuracy of 1 mm [4]. One MTx sensor module was attached to the assembly of the three coils to measure its orientation with respect to the global reference system. Before testing, the alignment between the orientation of the sensors and the laboratory was determined in order to express the signals from both systems in the same frame.

In the first experiments, the bench-test, the set of coils was placed on a table. One sensor was moved by hand near the coils. Distances were varied slowly from approximately 10 to 80 cm and the sensor was rotated along all axes. In the following experiments, the three perpendicular coils were attached to the body as illustrated in Fig. 1. One sensor was placed on the back of a subject, at the level of the first thoracic vertebra and one sensor was placed on the upper arm. The subject performed flexion–extension and abduction–adduction of the arm followed by standard anatomical movements of the back: flexion (and extension), lateral flexion and rotation. In the final tests, the sensor was placed on the upper leg, just above the knee. The subject walked across the laboratory at a comfortable pace for a number of steps. All experiments were repeated 10 times.

4. Experimental results

Fig. 5 shows the magnetic pulses as measured by the magnetometers from a typical trial. The sequence of the three X, Y and Z-pulses, as illustrated in Section 2, can be identified as well as their changes in magnitude as a result of the performed movement. The pulsewidth was 60 ms, the cycle time (B_1 to B_1) was 600 ms and the current 1.5 A.

The time instants of the pulses are exactly known and the rise time of a pulse is much faster than changes in the earth magnetic field vector caused by movement. Therefore, by evaluating the values of the earth magnetic field just before and after the pulse and high-pass filtering, the magnitudes of the pulses can be obtained. However, due to movement within one cycle of three pulses, errors can occur. An example can be seen in Fig. 6; in the first burst of three pulses, the dc earth field component varies.

In Fig. 6, an example of the distance estimates R from the center of the coils to the magnetic sensor is presented where the sensor is moved by hand. The root mean square (rms) accuracy of this trial is 7.9 mm compared to the optical distance measurement. When the sensor was moved beyond a distance of approximately 80 cm, the SNR of the magnetic signal was too low to be used for relevant measurements.

Fig. 7 shows the X, Y and Z-coordinates of a trial in which the subject performed latero-flexion of the back twice. The spherical parameters (distance R, and tracking angles α and β) were transformed in Cartesian coordinates. The center of the coils is the origin of the magnetic frame which is aligned with the global frame using the orientation $\mathbf{\Phi}$ of the source. The stars (*) present the magnetic position measurements and the solid line, the reference coordinates. From the initial coordinates, we can find that the sensor located on the back is about 45 cm positioned above the source (Z-coordinate), 5 cm to right (Y-

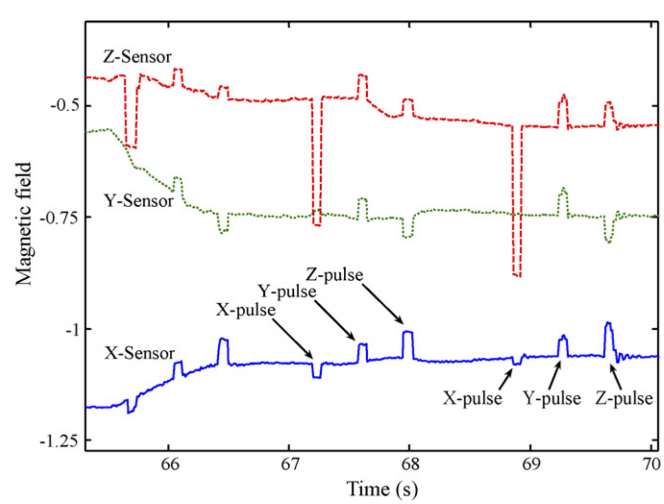

Fig. 5. Three cycles of magnetic pulses measured by the magnetometers. The pulsewidth was 60 ms and the cycle time 600 ms.

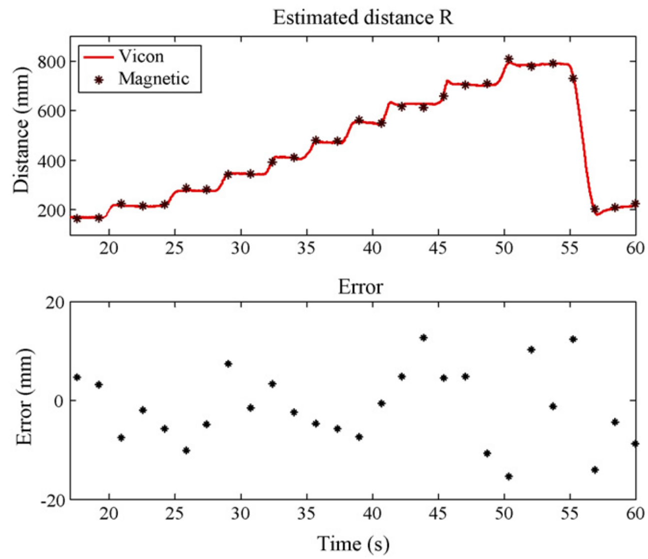

Fig. 6. Upper: magnetic and Vicon distance measurements of a typical bench-test recording. The sensor is moved by hand around the fixed source from approximately 15 to 80 cm. The magnetic estimates are indicated by stars (*). Lower: error in magnetic distance estimation compared to the reference Vicon.

Table 1
RMS Position and orientation errors and their standard deviations (S.D.) of the magnetic tracker for each segment and movement

Segment	Movement	Position error [mm]		Orientation error [°]	
		RMS	S.D.	RMS	S.D.
Bench-test		7.6	2.4	5.9	2.6
Back	Flexion	5.9	1.5	4.9	2.3
	Latero-flexion	6.3	1.5	5.2	2.1
	Rotation	5.9	1.4	5.1	1.9
	Walking	8.6	1.6	6.8	2.7
Arm	Flexion	7.8	1.8	6.6	2.2
	Abduction	7.2	1.7	6.2	2.2
	Walking	11.7	2.9	7.4	3.1
Leg	Walking	15.0	4.6	8.7	3.3

All movements were performed 10 times.

coordinate) and 6 cm forward (X-coordinate). First, the subject bends to the right and both the Y-coordinate and the Z-coordinate decrease. Then, the subject flexes through neutral position to the left. The Y-coordinate now increases while the Z-coordinate decreases. Since the sensor is placed on the right side of the spinal chord, the amplitude of the left flexion movement is less than the right. At the end of the two cycles, the subject is neutral position again. From the X-coordinate, we can see that the sensor moved only a few centimeters forward and backward during the recording.

Table 1 shows the numerical results of all performed experiments. The orientation error is defined as the smallest angle about which the sensor frame has to be rotated to coincide with the reference frame. The position error is defined as the shortest distance between the magnetic coordinates and the reference coordinates. The differences between the position and orientation measurements of the optical and magnetic system were taken from 10 trials for each movement. The lowest errors were observed during the relatively slow movements of the arm and back. The walking trials showed higher errors because of the under sampling with respect to the performed movement. The errors of the sensor on the back were smaller than those of the sensors on the leg or arm because during walking, the relative position of the sensor on the back was stable with respect to the source.

The orientation of the sensor module was also estimated using the signals from the gyroscopes, accelerometers and magnetometers and the fusion algorithm described in Ref. [20]. These estimates are independent of the distance between source and sensors, but they are correlated with accelerations and magnetic disturbances. The average orientation error using this method was 3.0° compared to orientation obtained using the optical reference system.

5. Discussion

In the experiments, we have demonstrated the feasibility of a portable magnetic tracker for human motion analysis. The performances do not yet meet requirements for some applications.

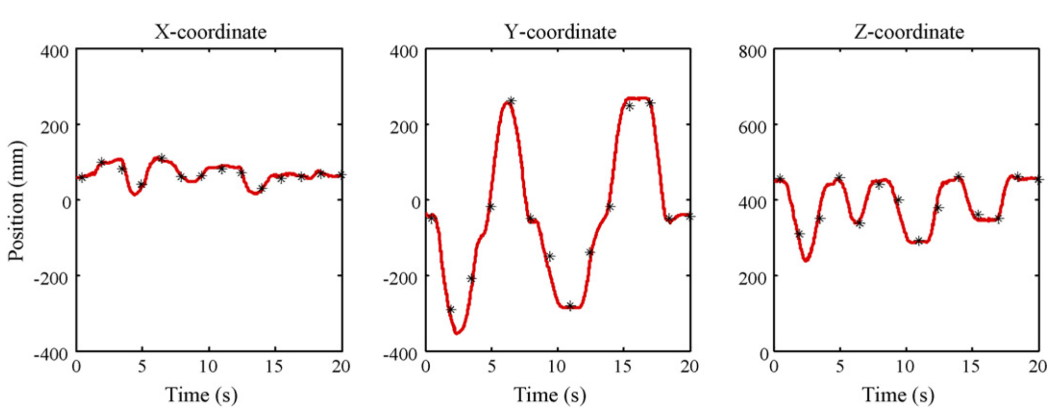

Fig. 7. X, Y and Z-coordinates of the sensor placed on the back with respect to the center of the source. The subject performed latero-flexion of the back twice. The magnetic position estimates are indicated by stars (*), the Vicon reference is the solid line.

A part of the errors was related to the low sampling frequency with respect to the performed movements. By combining the magnetic system with inertial sensors, we expect these errors to reduce. This combination will also improve the orientation estimates of the magnetic system.

The accuracy of the magnetic system can be improved by a higher signal-to-noise ratio, which will reduce the stochastic errors. This can be achieved by increasing the strength of the magnetic dipole or reducing the noise of the sensors. The configuration of the coils was optimized for distances up to 70 cm. This means that this set-up is not suitable for full body tracking. Paperno and Plotkin [18] found a significant improvement of the magnetic dipole approximation error of a coil by optimizing the length L of a coil with respect to its diameter b to an optimum of $L/b = 0.86$. A dipole strength necessary for distances on the body requires an optimal length that is not practical for body mounting. However, different coil configurations emitting stronger fields should be investigated. Also, a network of body attached coils (integrated in clothing) can be used for full body tracking. Systematic errors can be reduced by using the analytical relation to calculate the field emitted by a coil instead of the dipole approximation. However, it will require more processing time.

Although regular flipping of the magnetometers is necessary for stable magnetic measurements, it can introduce oscillations when pulsing due to large changes in the measured magnetic field. This can be seen in the last Z-pulse of Fig. 5. These measurement artifacts will decrease by reducing the frequency of flipping and by avoiding flipping during pulsing. The latter can be implemented by synchronizing it with the timing of actuation. The magnetoresistive sensors can saturate when placed in a large field, e.g. when the sensor is close to the coils. By controlling the current through the coil based on the strength of the measured field of the sensor, this effect can be minimized.

Within one burst of three magnetic pulses, the relative position and orientation between source and sensor can change. In the related 6 DOF calculation, these were assumed to be constant. To reduce these errors, the time between a X, Y and Z-pulse should be decreased. The pulse duration can be shortened within the specifications of the time constant of the coil. This also requires a higher sample frequency of the magnetometers. The changes in orientation and position of source and sensor during pulsing can be measured using inertial sensors.

The magnetic field magnitude decreases with the cube of distance. To measure a signal with a sufficient SNR, a strong field at the source is necessary and the sensors should be relatively close to the source. Continuous driving of all coils requires a substantial amount of energy. The tested update rate of this system was 1.7 Hz, which is low if this system is used for human motion tracking. However, this update rate is sufficient to serve as reference measurement for inertial tracking. Miniature inertial sensors are suitable for measuring fast changes in position and orientation and require less energy. Since the magnetic dipole-source is required to be active only during a small percentage of time, the average energy needed is limited. This principle requires an algorithm to fuse position and orientation estimates from the magnetic system with those of the inertial sensors. We hope to report on this shortly.

Like every magnetic tracking system, it is vulnerable for magnetic disturbances. The static magnetic field in the used measurement volume can be considered as homogeneous. The distance between source and sensors is relatively small, therefore less interference problems are expected compared to a range of several meters in a laboratory set-up. Moreover, by combining this system with inertial sensors, the effect of magnetic disturbances can be reduced. Errors related to magnetic disturbances will have different spatial and temporal properties than drift errors related to inertial sensors.

With the described settings and used batteries we were able to perform measurements for over 30 min. New generations of rechargeable batteries (or fuel cells) will have higher capacities which can extend the measurement time. Alternatively, the driving current I through the coils can be increased. In addition, shorter pulses and a longer cycle time when combined with inertial sensors will extent the operating time with a set of batteries.

6. Conclusion

In this study, a magnetic tracking device is presented that is ambulant and thus can fully be worn on the body without the need for an external reference. Although magnetic trackers are commercially available, they are limited to a restricted measurement volume and have large and heavy sources which do not allow for ambulatory purposes. We have designed a set of copper-winded coils, integrated in a synthetic dome and the electronics for battery powered magnetic pulsing. The accuracy of measurements is approximately 8 mm in position and 5° in orientation with 6 DOF movements. The system will be used as an aiding system to update on-body position and orientation estimates from inertial sensors. Although accuracy and mounting of the coils on a human body can be improved, it offers possibilities for ambulatory position and orientation measurements.

References

[1] E.B. Bachmann, Inertial and magnetic tracking of limb segment orientation for inserting humans into synthetic environments, PhD Thesis, Naval Postgraduate School, 2000.
[2] D.K. Cheng, Field and Wave Electromagnetics, Addison–Wesley, 1989.
[3] J.S. Day, G.A. Dumas, D.J. Murdoch, Evaluation of a long-range transmitter for use with a magnetic tracking device in motion analysis, J. Biomech. 31 (10) (1998) 957.
[4] Y. Ehara, H. Fujimoto, S. Miyazaki, M. Mochimaru, S. Tanaka, S. Yamamoto, Comparison of the performance of 3D camera systems II, Gait Posture 5 (1997) 251–255.
[5] S. Emura, S. Tachi, Multisensor integrated prediction for virtual reality, Presence 7 (4) (1998) 410–422.
[6] E. Foxlin, M. Harrington, G. Pfeifer, Constellation: a wide-range wireless motion-tracking system for augmented reality and virtual set applications, in: Proceedings of SIGGRAPH ’98, Orlando, Florida, 1998.
[7] D. Giansanti, V. Macellari, G. Maccioni, A. Cappozzo, Is it feasible to reconstruct body segment 3-D position and orientation using accelerometric data? IEEE Trans. Biomed. Eng. 50 (4) (2003) 476–483.
[8] D. Giansanti, V. Macellari, G. Maccioni, V. Macellari, The development and test of a device for the reconstruction of 3-D position and orienta-

tion by means of a kinematic sensor assembly with rate gyroscopes and accelerometers, IEEE Trans. Biomed. Eng. 52 (7) (2005) 1271–1277.
[9] K. Kobayashi, L. Gransberg, E. Knutsson, P. Nolen, A new system for three-dimensional gait recording using electromagnetic tracking, Gait Posture 6 (1) (1997) 63.
[10] J. Kuipers, Object tracking and determining orientation of object using coordinate transformation means, system and process, US Patent 3,868,565 (February 25, 1975).
[11] J. Kuipers, Tracking and determining orientation of object using coordinate transformation means, system and process, US Patent 3,983,474 (September 26, 1976).
[12] J.B. Kuipers, Quaternions and Rotation Sequences, Princeton University Press, 1999.
[13] L. LaScalza, J. Arico, R. Hughes, Effect of metal and sampling rate on accuracy of flock of birds electromagnetic tracking system, J. Biomech. 36 (2003) 141–144.
[14] H.J. Luinge, P.H. Veltink, Inclination measurement of human movement using a 3D accelerometer with autocalibration, IEEE Trans. Neural Syst. Rehabil. Eng. 12 (1) (2004) 112–121.
[15] H.J. Luinge, P.H. Veltink, C.T.M. Baten, Ambulatory measurement of arm orientation, J. Biomech. 40 (2007) 78–85.
[16] R.E. Mayagoitia, A.V. Nene, P.H. Veltink, Accelerometer and rate gyroscope measurement of kinematics: an inexpensive alternative to optical motion analysis systems, J. Biomech. 35 (4) (2002) 537–542.
[17] M.A. Nixon, B.C. McCallum, W.R. Fright, N.B. Price, The effect of metals and interfering fields on electromagnetic trackers, Presence 7 (2) (1998) 204–218.
[18] E. Paperno, A. Plotkin, Cylindrical induction coil to accurately imitate the ideal magnetic dipole, Sens. Actuators A: Phys. 112 (2–3) (2004) 248–252.
[19] F.H. Raab, E.B. Blood, T.O. Steiner, H.R. Jones, Magnetic position and orientation tracking system, IEEE Trans. Aerospace Electron. Syst. 15 (5) (1979) 709–718.
[20] D. Roetenberg, H.J. Luinge, C.T.M. Baten, P.H. Veltink, Compensation of magnetic disturbances improves inertial and magnetic sensing of human body segment orientation, IEEE Trans. Neural Syst. Rehabil. Eng. 13 (2005) 395–405.
[21] T. Stork, Electronic compass design using KMZ51 and KMZ52, Philips Semiconductors Application Note, AN00022, 2000.
[22] Xsens Technologies, MTx data sheet, http://www.xsens.com.
[23] R. Zhu, Z. Zhou, A real-time articulated human motion tracking using tri-axis inertial/magnetic sensors package, IEEE Trans. Neural Syst. Rehabil. Eng. 12 (2) (2004) 295–302.

Biographies

D. Roetenberg received the MSc degree in electrical engineering from the University of Twente in 2001. In 2002, he was with the Roessingh Research and Development, Enschede, The Netherlands, where he worked on EMG signal processing. In 2006, he received the PhD degree from the University of Twente on the topic of inertial and magnetic sensing of human motion. The work involved the development an ambulatory magnetic and inertial tracking system and sensor fusion algorithms. Currently, he is with Xsens Technologies B.V. where he works on human motion capture technology and aided inertial navigation solutions.

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