Prospective Motion Correction using Miniature Radio Frequency Coils

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Key Points

- The positions of miniature radio frequency (RF) coils inside the MRI scanner can be measured with high accuracy and precision, and in a temporal resolution suitable for real time MRI applications. A minimum of three such RF coils are required to uniquely define any arbitrary 3D rigid body motion.
- The RF coils may be implemented as "*wired* markers" where each RF coil is connected to the MRI scanner via a traditional coaxial cable.
- The RF coils may also be implemented as "*wireless* markers" that are free of any mechanical connection to the MRI scanner. Instead, the signal is wirelessly transmitted via inductive coupling to the nearby imaging (e.g. head) coil.
- The use of wired or wireless markers as position-tracking probes has been the foundation of several recent advances in prospective motion correction for brain MRI. The technique has been shown to significantly improve image quality of 2D/3D structural MRI, as well as increased the statistical significance of functional MRI.

Problem

Head motion is a fundamental problem in magnetic resonance imaging (MRI) of the brain. In structural MRI, motion artifacts such as blurring and ghosting can render images diagnostically unusable. In functional MRI, slight movements can result in large, artifactual, signal changes in the time-series data that will obscure the relatively small brain activation signals. These difficulties underline the importance of an effective motion correction strategy.

The prospective motion correction method presented here uses miniature radio-frequency (RF) coils as the fiducial for head motion tracking. The hardware for two different variants of the RF coils – namely the "wired marker" and "wireless marker" versions is described. The tracking pulse sequence and algorithm to measure the marker positions is discussed. Finally, the complete prospective motion correction package that utilizes these elements is reviewed, along with results obtained using this package for structural and functional brain MRI applications.

Miniature RF Coil Hardware

<u>Wired Markers</u>

Monitoring the positions of miniature RF coils using magnetic resonance was introduced in 1986 (1), and initially developed as a means of actively tracking the tip of a catheter or surgical tool during MRI guided interventional procedures (2). The method was later adapted for prospective motion correction applications for brain MRI (3,4), and has been the foundation for several recent advances in the field (5-14). We broadly refer to this family of RF coils (1-14) as "wired markers", since each RF coil is connected to the MRI scanner via a traditional coaxial cable.

The circuit schematics of example wired markers is shown in Figure 1a, b. Figure 1a shows a wired marker that is a tuned and matched resonant circuit. The resonant circuit is formed using a solenoid inductor (e.g. 3-turn, 4 mm diameter) and capacitor network, is tuned to the ^{1}H Larmor frequency and matched to the 50 Ω receiver. A PIN diode actively detunes the resonant circuit during RF transmit.

Figure 1b shows a wired marker that is untuned, consisting of the solenoid inductor without any capacitors. Despite the lower signal to noise ratio (SNR) of the untuned vs. tuned wired marker, it has been shown that the signal from the untuned marker is sufficient for position tracking (7). An untuned marker possesses the benefits of a simplified, smaller scale wired marker design, and also reduces the risk of RF induced heating, and image artifact in the event of insufficient RF blocking during RF transmit.

Inside the cavity of both solenoids is a small glass sphere (3 mm diameter) filled with a Gd-doped water solution. The spherical sample is the point source that is tracked. The spherical symmetry minimizes orientation dependence, and its short T_1 allows full signal recovery in the presence of multiple RF-pulse excitations. The base of each wired marker may also be RF shielded to reduce signal from the patient.

Each wired marker is attached via a coaxial cable to a multi-connect "interface box" (not shown), which is typically supplied by the MRI vendor. The interface box contains the necessary circuitry to connect a custom RF coil to the MRI scanner, and includes pre-amplifiers, matched networks, DC bias circuitry for active detuning, and RF coil identification. The interface box then connects to the MRI scanner's standard front-end coil connector. For simultaneous imaging and motion tracking, the MRI scanner should be equipped with a number of receiver channels≥ number of channels required for the imaging coil + number of wired markers.

Figure 1c shows a subject wearing a headband containing three wired markers, which serves as the fiducial for head motion tracking (5,6,12,14). Three wired markers are used since a minimum of three points are required to uniquely define any arbitrary 3D rigid body motion.

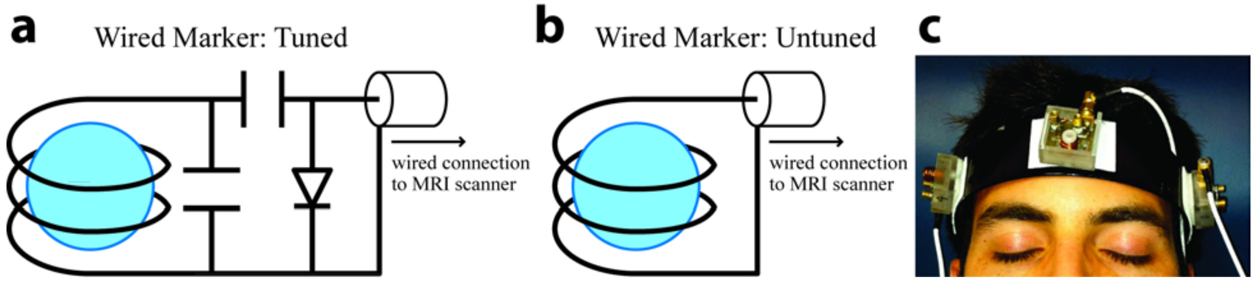

Figure 1. Circuit diagrams of example wired markers. a) Tuned wired marker (5). b) Untuned wired marker (7). c) Headband containing three tuned wired markers for prospective motion correction of brain MRI (5,6,12,14).

<u>*Wireless Markers*</u>

Inductively coupled RF coils have been previously used for local image SNR improvement (15), as well as visualization of fiducials (16), stents (17), and catheters (18). Position tracking by attaching these RF coils to MRI guided surgical devices has also been demonstrated using image based methods (19,20). Most recently, studies have introduced inductively coupled miniature RF coils for prospective motion correction of brain MRI (21,22). We refer to this variant of RF coils as "wireless markers", since they are free of any mechanical connection to the scanner. Wireless markers possess several advantages vs. wired markers, including eliminating the RF safety risks from electrical cables, reduced tracking errors due to tugging on cables, reduced hardware load since no additional receiver channels or interface circuitry is required, and reduced setup time.

The circuit schematic of a wireless marker is shown in Figure 2a (bottom). It consists of a solenoid inductor (e.g. 3-turn, 4 mm diameter) and capacitor that is resonant at the ^{1}H Larmor frequency. Inside the solenoid cavity is a small glass sphere (3 mm diameter) filled with Gd-doped water solution that is identical to Figure 1. A fast-switching crossed diode passively detunes the resonant circuit during RF transmit. Figure 2b shows a picture of a single wireless marker. Figure 2c shows a subject wearing a pair of glasses integrated with three wireless markers for prospective motion correction of brain MRI. (21)

In the absence of any cable connections, wireless marker signal transmission is achieved by inductively coupling the wireless markers to the nearby imaging (e.g. head) coil. This is illustrated in Figure 2a. During RF receive, each wireless marker acts as a local signal amplifier that picks up the MR signal in its immediate vicinity, which is dominated by the spherical sample. The signal generates a current *di/dt* in the wireless marker, and an associated magnetic field (dashed lines) and flux *dΦ/dt* as it passes through the imaging coil. The flux induces a voltage *V* in the imaging coil according to Faraday's law: $di/dt \propto d\Phi/dt \propto V$, which is then routed to the MRI scanner's standard RF receiver. The key concept here is that even though these two RF coils are not physically connected, the signal from the wireless marker is transmitted to the imaging coil via the magnetic flux *dΦ/dt* that links the two coils.

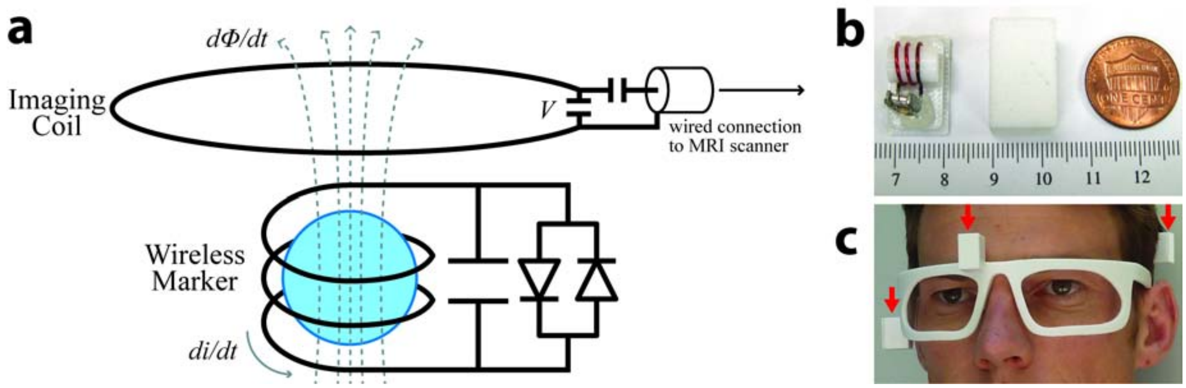

Figure 2[*]. a) Circuit diagram of a wireless marker (bottom) and illustration of wireless signal transmission via inductive coupling with a nearby imaging coil (top). b) Wireless marker (left), enclosing capsule (middle), and U.S. penny (right), relative to ruler markings in mm. c) Polycarbonate glasses integrated with three wireless markers for prospective motion correction of brain MRI. [[*]Figure reproduced from ref. (21)]

Tracking Pulse Sequence

The tracking pulse sequence (2) used to measure the 3D positions of either wired or wireless markers is shown in Figure 3a. A non-selective RF pulse ($\alpha = 1°$) is followed by gradient echo readouts along the physical x, y, and z gradient axes, resulting in three 1D projections along orthogonal axes. The low flip angle of the RF excitation minimizes the effect of the tracking pulse sequence on the imaged object. Spoiler gradients dephase the magnetization in large volumes (from the subject) while preserving signal from the smaller spherical samples inside each marker. Alternative tracking pulse sequences such as Hadamard encoding schemes may also be employed (2,23).

Figure 3b shows the signals received from three wired markers after a single 1D projection. The "wired signals" S_1, S_2, S_3 (Figure 3b, left) are obtained from receiver channels 1, 2, 3, which are connected via coaxial cable to wired markers 1, 2, 3, respectively. The wired signals S_1, S_2, S_3 are color coded red, green, and blue to illustrate that the signal originating from each wired marker can be separately and unambiguously identified, since each wired marker is directly connected to its own independent receiver channel. Peaks are clearly visible at three locations, which correspond to the positions of the three wired markers along the projected axis. Peak searches in all three 1D projections therefore yields each wired marker's 3D coordinates.

It is important to note that, in addition to the wired signals S_1, S_2, S_3, the three wired markers will also generate a "wireless signal" S_{img} that can be obtained at the imaging coil receiver channel (Figure 3b, right). This wireless signal S_{img} shows the signals from the three wired markers as they inductively couple to the imaging coil. In the case of three wireless markers, S_{img} is the only measurement available for calculating the marker positions. From S_{img} alone it is not immediately obvious which peak corresponds to which marker, since the signals from all three markers are simultaneously coupled to the imaging coil.

This unknown peak-to-marker assignment is referred to as the “correspondence problem”. This problem can be solved by using *a priori* knowledge of the markers’ geometrical arrangement, and by selecting a predefined geometrical arrangement that maximizes the separation of the markers in x, y, z, such that the peaks will not overlap for the motions in a typical scan (21). A solution has also been proposed that involves a brute force search of all possible marker locations to find the polygon that best matches the geometrical arrangement of the actual markers (22). Peak searches in all three 1D projections, together with the solution to the correspondence problem, will yield each wireless marker’s 3D coordinates.

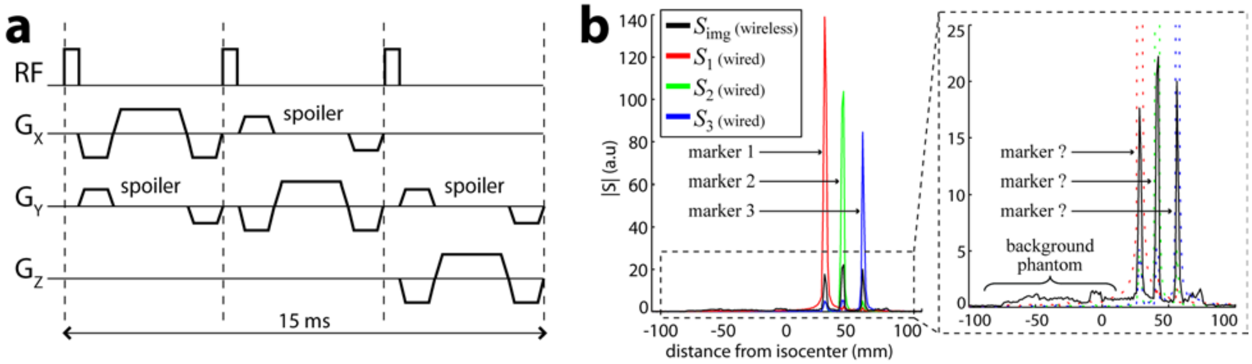

Figure 3. a) Tracking pulse sequence used to measure the 3D positions of either wired or wireless markers. b) Signal from three wired markers after a single 1D projection. The “wired signals” S_1, S_2, S_3 are obtained from receiver channels 1, 2, 3, which are directly connected to markers 1, 2, 3 via coaxial cable. The “wireless signal” S_{img} is obtained from the imaging coil receiver channel and shows the signals from the three wired markers as they inductively couple to the imaging coil.

Prospective Motion Correction

Prospective correction for rigid head motions, using the real-time tracking data from either wired or wireless markers, can be flexibly interleaved into a variety of imaging sequences due to the relatively short temporal footprint (~ 15 ms) of the tracking pulse sequence.

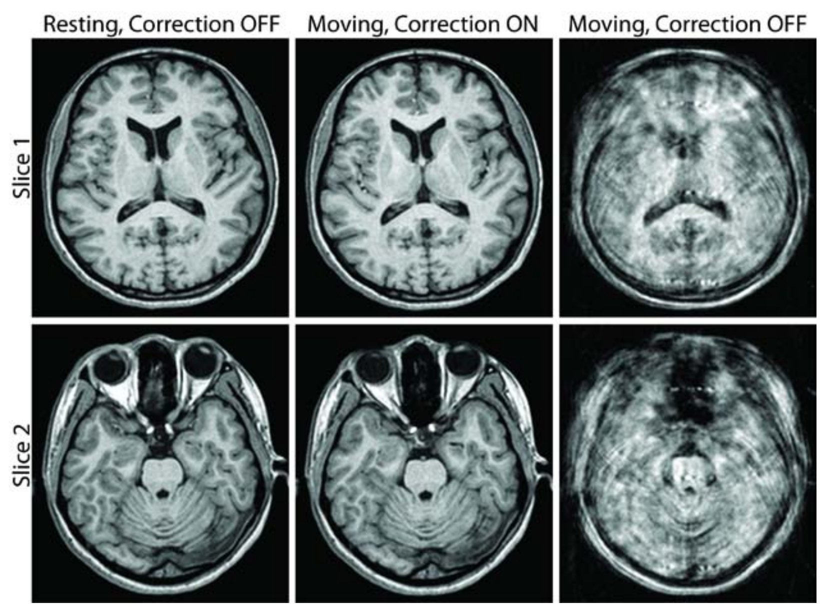

Figure 4. Prospective motion correction of a 3D-MPRAGE scan using the wired markers in Figure 1c. During the scans in columns 2, 3, the volunteer performed continuous, reproducible, ± 5° left-right head rotations (i.e. head shaking “no”) throughout the entire scan.

A flowchart of a typical prospective motion correction strategy is shown in ref. (5) Figure 2. The tracking pulse-sequence (Figure 3a) is inserted into the imaging sequence between every user defined number of imaging phase-encode(s), thereby continually measuring the marker positions throughout the scan. For each position measurement, the six degrees of freedom rigid transform is calculated (24) that realigns the current marker positions to the original (reference) positions at the beginning of the scan. This transform is then applied to update the image-volume orientation and position before the next imaging phase-encode(s) are acquired.

The technique has been shown to improve the image quality of 2D/3D structural MRI scans (e.g. GE, SPGR, MPRAGE, FSE) acquired during subject motion (5,21,22). The technique is also compatible with EPI-based scans, and improved statistical significance has been shown in functional MRI studies with subjects performing deliberate motions (6,12) as well as with subjects instructed to remain still (14).

Figure 4 shows an example of prospective motion correction using wired markers on a 3D-MPRAGE scan. Figure 5 shows prospective motion correction using wireless markers on a 2D-FSE scan. In both examples brain images are shown for the resting, motion-corrected, and motion-corrupted datasets. Without correction, images are corrupted by motion artifacts (column 3) such as blurring and ghosting. Prospective correction (column 2) results in virtually perfect correction relative to the resting images (column 1), with fine edges and details of anatomical structures being well preserved.

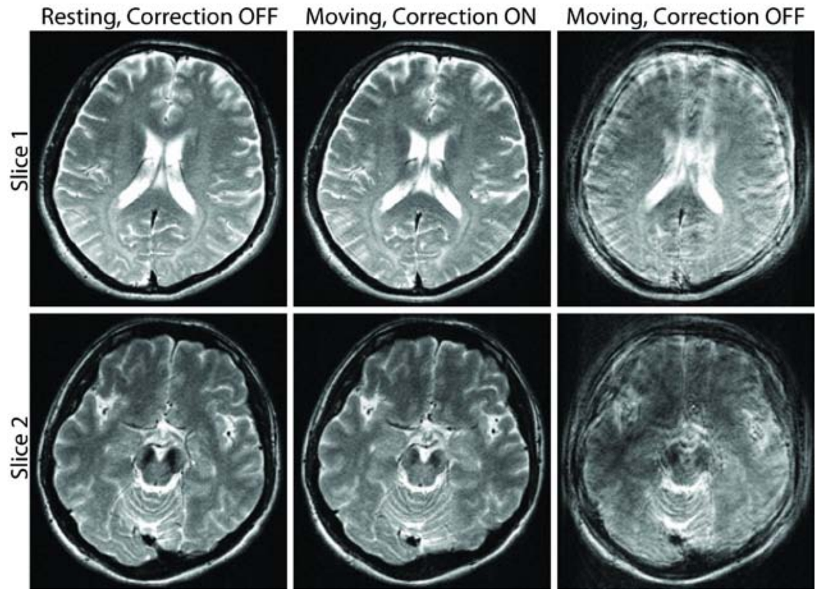

Figure 5. Prospective motion correction of a 2D-FSE scan using the wireless markers in Figure 2c. During the scans in columns 2, 3, the volunteer performed abrupt, reproducible, ± 5° head rotations in three orthogonal directions (i.e. head shaking "no", nodding "yes", and tilting from side-to-side) at 45 sec intervals throughout the entire scan.

Summary

- Miniature RF coils for position tracking inside the MRI scanner may be implemented using either a wired (tuned/untuned) or wireless marker schematic. Wired markers are directly connected to the MRI scanner's standard receiver chain via coaxial cables. Wireless markers are free of any mechanical connection to the MRI scanner, and instead transmit their signal via inductive coupling with a nearby imaging coil.
- A tracking pulse sequence is used to measure the 3D positions of the markers. In the case of wireless markers, an additional peak-to-marker correspondence problem must also be solved.
- Prospective motion correction using either wired or wireless marker tracking may be incorporated into a variety of structural and functional imaging sequences.

References

1. Ackerman JL, Offutt MC, Buxton RB, Brady TJ. Rapid 3D Tracking of Small RF Coils. Proceedings of the 5th Annual Meeting of SMRM. Montreal, QC, Canada1986. p 1131-1132.
2. Dumoulin CL, Souza SP, Darrow RD. Real-time position monitoring of invasive devices using magnetic resonance. Magnetic Resonance in Medicine 1993;29(3):411-415.
3. Derbyshire JA, Wright GA, Henkelman RM, Hinks RS. Dynamic scan-plane tracking using MR position monitoring. Journal of Magnetic Resonance Imaging 1998;8(4):924-932.
4. Krueger S, Schaeffter T, Weiss S, Nehrke K, Rozijn T, Boernert P. Prospective Intra-Image Compensation for Non-Periodic Rigid Body Motion Using Active Markers. Proceedings of the 14th Annual Meeting of ISMRM. Seattle, WA, USA2006. p 3196.
5. Ooi MB, Krueger S, Thomas WJ, Swaminathan SV, Brown TR. Prospective real-time correction for arbitrary head motion using active markers. Magn Reson Med 2009;62(4):943-954.
6. Ooi MB, Krueger S, Muraskin J, Thomas WJ, Brown TR. Echo-planar imaging with prospective slice-by-slice motion correction using active markers. Magn Reson Med 2011;66(1):73-81.
7. Ooi MB, Aksoy M, Watkins RD, Bammer R. High Precision Tracking of Un-Tuned Micro-Coils for Real-Time Motion Correction Applications. Proceedings of the 20th Annual Meeting of ISMRM. Melbourne, Australia2012. p 3428.
8. Aksoy M, Ooi MB, Watkins RD, Kopeinigg D, Forman C, Bammer R. Combining Active Markers and Optical Tracking for Prospective Head Motion Correction. Proceedings of the 20th Annual Meeting of ISMRM. Melbourne, Australia2012. p 3430.
9. Beall EB, Lowe MJ. An efficient EPI pulse sequence module for active marker motion correction acquisition for EPI scans. Proceedings of the 20th Annual Meeting of ISMRM. Melbourne, Australia2012. p 4176.
10. Haeberlin M, Kasper L, Brunner DO, Barmet C, Pruessmann KP. Continuous Motion Tracking and Correction Using NMR Probes and Gradient Tones. Proceedings of the 20th Annual Meeting of ISMRM. Melbourne, Australia2012. p 595.
11. Sengupta S, Gore JC, Welch EB. Prospective motion correction using NMR probes. Proceedings of the 20th Annual Meeting of ISMRM. Melbourne, Australia2012. p 2468.
12. Ooi MB, Muraskin J, Zou X, Thomas WJ, Krueger S, Aksoy M, Bammer R, Brown TR. Combined prospective and retrospective correction to reduce motion-induced image misalignment and geometric distortions in EPI. Magn Reson Med 2013;69(3):803-811.
13. Qin L, Schmidt EJ, Tse ZT, Santos J, Hoge WS, Tempany-Afdhal C, Butts-Pauly K, Dumoulin CL. Prospective motion correction using tracking coils. Magn Reson Med 2013;69(3):749-759.
14. Muraskin J, Ooi MB, Goldman RI, Krueger S, Thomas WJ, Sajda P, Brown TR. Prospective active marker motion correction improves statistical power in BOLD fMRI. Neuroimage 2013;68:154-161.
15. Schnall MD, Barlow C, Subramanian VH, Leigh Jr JS. Wireless implanted magnetic resonance probes for in vivo NMR. Journal of Magnetic Resonance (1969) 1986;68(1):161-167.
16. Burl M, Coutts GA, Young IR. Tuned fiducial markers to identify body locations with minimal perturbation of tissue magnetization. Magn Reson Med 1996;36(3):491-493.
17. Quick HH, Kuehl H, Kaiser G, Bosk S, Debatin JF, Ladd ME. Inductively coupled stent antennas in MRI. Magn Reson Med 2002;48(5):781-790.
18. Quick HH, Zenge MO, Kuehl H, Kaiser G, Aker S, Massing S, Bosk S, Ladd ME. Interventional magnetic resonance angiography with no strings attached: wireless active catheter visualization. Magn Reson Med 2005;53(2):446-455.
19. Flask C, Elgort D, Wong E, Shankaranarayanan A, Lewin J, Wendt M, Duerk JL. A method for fast 3D tracking using tuned fiducial markers and a limited projection reconstruction FISP (LPR-FISP) sequence. J Magn Reson Imaging 2001;14(5):617-627.
20. Busse H, Trampel R, Grunder W, Moche M, Kahn T. Method for automatic localization of MR-visible markers using morphological image processing and conventional pulse sequences: feasibility for image-guided procedures. Journal of magnetic resonance imaging : JMRI 2007;26(4):1087-1096.
21. Ooi MB, Aksoy M, Maclaren J, Watkins RD, Bammer R. Prospective motion correction using inductively coupled wireless RF coils. Magn Reson Med 2013.
22. Sengupta S, Tadanki S, Gore JC, Welch EB. Prospective real-time head motion correction using inductively coupled wireless NMR probes. Magn Reson Med 2013.
23. Dumoulin CL, Mallozzi RP, Darrow RD, Schmidt EJ. Phase-field dithering for active catheter tracking. Magn Reson Med 2010;63(5):1398-1403.
24. Umeyama S. Least-squares estimation of transformation parameters between two point patterns. IEEE Transactions on Pattern Analysis and Machine Intelligence 1991;13(4):376-380.