Three-dimensional Optical Flow Method for Measurement of Volumetric Brain Deformation from Intraoperative Magnetic Resonance Images

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Short title: Brain shift measurement by optical flow

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Abstract

A three-dimensional optical flow method to measure volumetric brain deformation from sequential intraoperative Magnetic Resonance Images (MRIs) and preliminary clinical results from five cases are reported. Intraoperative MRIs were scanned before and after dura opening, twice during tumor resection, and immediately after dura closure. The maximum cortical surface shift measured was 11 mm and subsurface shift was 4 mm. The computed deformation field was most satisfactory when the skin was segmented and removed from the images before the optical flow computation.
Introduction

Intraoperative brain shift is a problematic issue in image guided neurosurgery, since any spatial discrepancy between coregistered preoperative images and the shifted brain diminishes the accuracy of neuronavigation (1-4).

For more accurate and reliable image-guided navigation, the preoperative image should be updated to reflect intraoperative changes in the shape of the brain by deformable registration of pre- to intraoperative images. This enables the incorporation of high-resolution physiologic and anatomical information in the preoperative images, which are not otherwise available.

This paper will introduce our approach to assess intracranial structural change by optical flow measurement. Intraoperative images of brain deformation were obtained in an intraoperative magnetic resonance imaging system (Signa SP, 0.5T, General Electric Medical Systems, Milwaukee, WI), which has two toroidal magnetic coils and allows access to the surgical field through a 56-cm gap between the coils.
Methods

Optical flow

We developed a method to measure brain deformation from intensity change in sequential MRIs. Deformation is estimated with a three-dimensional optical flow measurement based on local intensity differences. A multiresolution approach was used to efficiently estimate the deformation. The method was originally developed for two-dimensional image measurement, and we modified it to process volumetric images.

Optical flow, a method for computing a motion field from images, has been employed mostly in the computer vision and artificial intelligence community since the late 1970s. By definition, optical flow is an intensity-based approximation to image motion from sequential time-ordered images. Optical flow has been applied to motion detection, segmentation and motion-compensated image encoding. Detailed comparison of various optical flow methods appears in Barron et al.'s report (5).

The underlying assumption of optical flow is that the image intensity $E$ of a moving point $(x, y, z)$ at a time point $t$ is constant after a short duration of time $\delta t$. If a vector $(u, v, w)$ represents the velocity of the point and the intensity of the point does not change over the time $\delta t$, we can formulate,

$$E(x + u\delta t, y + v\delta t, z + w\delta t, t + \delta t) = E(x, y, z, t).$$  

(1)
Assuming that the image intensity varies smoothly with \(x,y,z\) and \(t\), we can use a first-order Taylor expansion of the left hand side of Eq. (1):

\[
E(x,y,z,t) + u \frac{\partial E}{\partial x} + v \frac{\partial E}{\partial y} + w \frac{\partial E}{\partial z} + \frac{\partial E}{\partial t} = E(x,y,z,t). \tag{2}
\]

Dividing by \(\delta t\), we obtain:

\[
\frac{\partial E}{\partial x} u + \frac{\partial E}{\partial y} v + \frac{\partial E}{\partial z} w + \frac{\partial E}{\partial t} = 0. \tag{3}
\]

In this paper, we abbreviate the first-order derivatives by using subscripts. So Eq. (3) is written with the abbreviations \(E_x = \frac{\partial E}{\partial x}\), \(E_y = \frac{\partial E}{\partial y}\) and \(E_z = \frac{\partial E}{\partial z}\), as:

\[
E_x u + E_y v + E_z w + E_t = 0. \tag{3'}
\]

Eq. (3') is the optical flow constraint equation to solve for \((u,v,w)\). However, as in two-dimensional cases, we cannot solve \((u,v,w)\) in the direction of the iso-brightness contours, or perpendicular to the intensity gradient (details in (6) "the aperture problem").

**Formulation of energy function**

We followed the approach of Horn and Schunck (6) for solving the underdetermined optical flow problem based on the gradient of the images. We construct a total error measure \(E(\alpha)\) as the
spatial integral of two terms; the first term $E_o = (E_x u + E_y v + E_z w + E_t)^2$ is the square of the error from the optical flow constraint Eq. (3), and the second term $E_s = u_x^2 + u_y^2 + u_z^2 + v_x^2 + v_y^2 + v_z^2 + w_x^2 + w_y^2 + w_z^2$ is a penalty that encourages smoothness:

$$E(\alpha) = \int\int\int \left( E_o(x, y, z, t) + \alpha^2 E_s(x, y, z, t) \right) dx dy dz. \quad (4)$$

$\alpha^2$ is a relative weight of the two terms determined by considering the signal-to-noise ratio of the second term $E_s$.

The smoothness constraint $E_s$ was derived from Horn and Schunck’s smoothness constraint in the two-dimensional case (6). The assumption here is that neighboring points on the objects have similar displacements. We may solve (4) by the calculus of variations (7). The Euler-Lagrange equations yield:

$$\nabla^2 u = \frac{1}{\alpha^2} (E_x u + E_y v + E_z w + E_t) E_x$$

$$\nabla^2 v = \frac{1}{\alpha^2} (E_x u + E_y v + E_z w + E_t) E_y$$

$$\nabla^2 w = \frac{1}{\alpha^2} (E_x u + E_y v + E_z w + E_t) E_z. \quad (5)$$

where $\nabla^2$ is the Laplacian operator

$$\nabla^2 = \frac{\partial^2}{\partial x^2} + \frac{\partial^2}{\partial y^2} + \frac{\partial^2}{\partial z^2}.$$
We approximate the Laplacians of $u$, $v$, and $w$ by

\[ \nabla^2 u \approx \bar{u}_{i,j,k} - u_{i,j,k} \]

\[ \nabla^2 v \approx \bar{v}_{i,j,k} - v_{i,j,k} \quad (6) \]

\[ \nabla^2 w \approx \bar{w}_{i,j,k} - w_{i,j,k} \]

where $\bar{u}_{i,j,k}$, $\bar{v}_{i,j,k}$ and $\bar{w}_{i,j,k}$ denote Gaussian weighted average ($\sigma = 0.75$) of the neighbor points around the point at $(i, j, k)$.

Combining Eq. (5) with Eq. (6):

\[ \alpha - u = \frac{1}{\alpha^2} ( E_x u + E_y v + E_z w + E_t ) E_x \]

\[ \alpha - v = \frac{1}{\alpha^2} ( E_x u + E_y v + E_z w + E_t ) E_y \quad (7) \]

\[ \alpha - w = \frac{1}{\alpha^2} ( E_x u + E_y v + E_z w + E_t ) E_z \]

and isolating $u$, $v$, and $w$ leads to

\[ u = \bar{u} - \frac{E_x ( E_x \bar{u} + E_y \bar{v} + E_z \bar{w} + E_t )}{\alpha^2 + E_x^2 + E_y^2 + E_z^2} \]

\[ v = \bar{v} - \frac{E_y ( E_x \bar{u} + E_y \bar{v} + E_z \bar{w} + E_t )}{\alpha^2 + E_x^2 + E_y^2 + E_z^2} \quad (8) \]
Eq. (8) is a system of linear equations with about $10^8$ variables. Direct matrix methods such as Gauss-Jordan are computationally expensive, so we use the iterative Gauss-Seidel method (8).

According to this method, the updated $(u^{n+1}, v^{n+1}, w^{n+1})$ is computed from $(\overline{u}^n, \overline{v}^n, \overline{w}^n)$ with the following iteration:

$$w^{n+1} = \overline{w}^n - \frac{E_x (E_x \overline{u}^n + E_y \overline{v}^n + E_z \overline{w}^n + E_t)}{\alpha^2 + E_x^2 + E_y^2 + E_z^2}$$

$$v^{n+1} = \overline{v}^n - \frac{E_y (E_x \overline{u}^n + E_y \overline{v}^n + E_z \overline{w}^n + E_t)}{\alpha^2 + E_x^2 + E_y^2 + E_z^2}$$

$$u^{n+1} = \overline{u}^n - \frac{E_z (E_x \overline{u}^n + E_y \overline{v}^n + E_z \overline{w}^n + E_t)}{\alpha^2 + E_x^2 + E_y^2 + E_z^2}$$

$E_x$, $E_y$, and $E_z$ are approximations to the derivative and computed by averaging finite differentials in two input images from intraoperative MRI, namely a reference image of the brain before deformation and a target image after deformation, and applying Gaussian smoothing ($\sigma = 0.75$ pixels). We also approximated partial derivative of intensity over time $E_t$ by subtracting the reference from target image and applying Gaussian smoothing ($\sigma = 0.75$ pixels).

In our experiments, the images are resampled to be isotropic to have $1^3\text{mm/voxel}$. 
We employed a hierarchical multiresolution scheme to accommodate different scales of motion with computational efficiency. The multiresolution approach speeds up the convergence of the iteration represented by Eq. (9). We also expect that the multiresolution approach captures large-scale deformation in the low-resolution steps and small deformation in the fine-resolution step. The resolution ratios from the original data are x1/8, x1/4, x1/2, and x1, by downsampling by a factor of two in the slice direction of the image and smoothing by a Gaussian filter kernel.

After 500 iterations in each resolution, or when convergence is observed, the estimated motion field is passed to the next resolution level by bilinear interpolation and used as the next initial guess. $\alpha^2$ was 0.1 in the experiments presented in this paper.

Before applying the optical flow measurement, we initially registered the images according to scan locations encoded in the image headers. Although the patients’ heads were rigidly fixed by a modified MR-compatible Mayfield clamp (Ohio Medical Instruments, Cincinnati, OH), we could observe displacements of the head position mostly due to surgical maneuvers. To accommodate this motion, we performed an additional rigid registration by maximization of mutual information (MMI), which is described in detail by Viola and Wells et al. (9-11). This method automatically and directly works on medical images, in contrast to other methods that require the setting of fiducial makers or other manual interaction for registration.
Results

Phantom study

We validated the accuracy of the method using MRI scans of a deformable ultrasound gel pad (100x100x50 mm.). The accuracy testing was possible by comparing the deformation measured by the three-dimensional optical flow with actual measurement of landmark movement from the MR images.

The phantom had a hole (50mm in diameter) between the square faces to imitate a ventricle in brains. Twenty-seven wood sticks (approximately 2mm in diameter) were inserted parallel to the hole as landmarks. The phantom was placed between two upright plates which were 52mm apart and glued to a base stage and a weight (520g) was then placed on the top of the gel to press and deform the phantom.

After placing the phantom in the intraoperative 0.5T MR scanner, a three-dimensional volume of SPGR (spoiled gradient echo recovery) with 28 slices (TR: 28.6ms, TE: 12.8ms, FOV 170x170 mm, matrix 256x128, 1 NEX, 2 mm thickness/0 spacing) was obtained before and after the weight was applied. These images were processed by the three-dimensional optical flow method to predict the movement of the landmarks. We also performed the actual measurement of landmark displacement by collecting the landmark coordinates from the images.

In Figure 1, the images of the gel phantom and measurement results are displayed. The predicted landmark displacements by the three-dimensional optical flow matches with the actual
displacement of the landmark with an average error of 3.5 mm. The maximum displacement of the landmark was 20.0 mm and minimum was 1.0 mm, which are typical of actual brain deformation in surgeries.

Figure 1: (Top left) A MR image of ultrasound gel phantom taken by the 0.5T intraoperative MR scanner. The phantom has a hole (50-mm in diameter) in the middle
and 27 landmarks (wood sticks, 2-mm in diameter) appeared in the MRI as low intensity dots. (Top right) MR image of the phantom pressed by a weight from the top. The three-dimensional optical flow method was applied to these MR images and displacement of voxels in the images was predicted. Red arrows in the (bottom left) image indicates the in-plane components of displacement vectors. In the (bottom right) image, predicted displacement vectors of the landmarks (red arrows) and the actual displacement of the landmarks (blue arrows) matched reasonably well.

Clinical studies

We also applied the method to five sets of clinical data both for the evaluation of the method and for quantitative and qualitative evaluation of brain deformation. Profiles of the clinical cases are listed in Table 1 with the results from experimental evaluation tests.

In each case three-dimensional SPGR (spoiled gradient echo recovery) with 60 slices (TR: 28.6ms, TE: 12.8ms, FOV 240x240 mm, matrix 256x128, 1 NEX, 2.5 mm thickness/0 spacing) was obtained before and after dura opening, twice during tumor resection, and after dura closure (a total of five series per case). Note that all images are acquired during surgery in the interventional MRI scanner.
**Volume change experiment**

The first part of the test with the clinical data is a volume change test, in which change was measured by comparing the sequence of intraoperative MRIs right after the opening of the dura (namely the *reference* data) with the image acquired at the beginning of tumor resection (namely the *target* data). We also measured the volume change after motion correction of the *target* image. The motion correction was made possible by application of the inverse deformation field (indicating displacement from the *target* data to the *reference* data) to the *target* image. We subtracted aligned *reference* images from *target* images in order to measure mismatched volume based on the intensity difference. From the subtraction images, we counted voxels with more than a predefined intensity threshold computed from mean image intensity value of gray matter. This mean image intensity was computed from ten rectangular (10x10 pixels) representative regions selected manually from the *reference* images.

The computational setting for this test was an Ultra HPC 6000 (Sun Microsystems, Mountain View, CA, 20x 250 MHz CPUs). The computation time was approximately 17 minutes for pre-processing and deformation measurement; 500 iterations were performed in each resolution step. The result of the test is compiled in Table 1. The average volume change was 10,385 mm$^3$, and motion correction reduced the volume difference to 703 mm$^3$. From this we can conclude that the optical flow measurement can account for about 90% of volume change from sequential intraoperative MRIs.
Brain deformation analysis

We performed clinical feasibility tests in all of the five cases listed in Table 1 with qualitative and quantitative analysis of the deformation. Five pairs of image sets are analyzed: (1st) image scanned before dura opening vs. after dura opening, (2nd and 3rd) before dura opening vs. during tumor resection, (4th) before dura opening vs. right after dura closure. This section presents case #4, which has illustrative findings.

Table 1: Case list and volume discrepancy before/after motion correction

<table>
<thead>
<tr>
<th>Case number</th>
<th>History</th>
<th>Age</th>
<th>Sex</th>
<th>Before correction</th>
<th>After correction</th>
<th>Reduction ratio (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>#1</td>
<td>Lt. frontal mass</td>
<td>47</td>
<td>M</td>
<td>9995</td>
<td>299</td>
<td>97</td>
</tr>
<tr>
<td>#2</td>
<td>Rt. frontal mass</td>
<td>49</td>
<td>F</td>
<td>3633</td>
<td>145</td>
<td>96</td>
</tr>
<tr>
<td>#3</td>
<td>Posterior fossa mass</td>
<td>19</td>
<td>M</td>
<td>1679</td>
<td>411</td>
<td>76</td>
</tr>
<tr>
<td>#4</td>
<td>Rt. parietal mass</td>
<td>53</td>
<td>M</td>
<td>25055</td>
<td>1277</td>
<td>95</td>
</tr>
<tr>
<td>#5</td>
<td>Rt. frontal mass</td>
<td>36</td>
<td>F</td>
<td>11564</td>
<td>1387</td>
<td>88</td>
</tr>
<tr>
<td>Average</td>
<td></td>
<td></td>
<td></td>
<td>10385</td>
<td>703</td>
<td>90</td>
</tr>
</tbody>
</table>

The motion correction was made possible by application of the inverse deformation field (indicating displacement from the target data to the reference data) to the target image. In the subtraction image from the (pre- and post-motion-corrected) reference and the target images, number of voxels with image intensity over median intensity value of gray matter were counted to approximate volume change.

In case #4, a metastasis in the right precentral gyrus (motor cortex) was removed. Intraoperatively, the cortex showed a consistent sinking, associated with cerebrospinal fluid
(CSF) drainage and tumor removal. The elongation of the vectors, as displayed in the original images accurately depicts this clinical finding (Figure 2). At present, we concentrate on the cortical shift to refine the computational method. Nevertheless, it is conceivable that the subsurface movement in the vicinity of the ventricles, away from the region of tumor resection, is adequately detected.

**Figure 2**: Sequential deformation vector map from case #4 (Left: after dura opening. Right: beginning of tumor resection.) Intraoperatively the cortex showed a consistent sinking, associated with cerebrospinal fluid (CSF) drainage and tumor removal. The elongation of the vectors is judged to accurately depict the clinical finding of a consistent sinking of the cortex. The subsurface movement in the vicinity of the ventricles, away
from the region of tumor resection, is also detected in the right image. Maximum cortical surface shift measured was 11 mm, and subsurface shift around the ventricle was 4 mm.

The maximum cortical surface shift measured was 11 mm; subsurface shift around the ventricle was 4 mm. Volume change was measured by aligning the sequence of intraoperative MRI before opening of the dura to the image acquired immediately after the dura was opened.

**Figure 2** shows sequential movements of randomly selected points along the gravity direction computed with the method. Two sites near the cortical surface and ventricle were selected from the image taken before the craniotomy; four points (6 mm apart) in each site were tracked by the three-dimensional optical flow. In Figure 3, the gravity-directional component of the displacement vectors shows local shifting near the cortex (maximum approximately -8 mm) and ventricle (approximately -2 mm). The global shift was observed in the tissue within approximately 10 mm of the surface layer. Notice the rebounding of the surface after the dura closure. These findings also match with the color mapping of the magnitude of the displacement vector compiled from the results of case #1 (Figure 4). The mapping indicates that the left frontal cortex shifts approximately 0 to 8 mm along the gravity direction.
Figure 3: The brain shift along the gravity direction by tracking four points (each 6 mm apart) at a cortical site (single arrow) and ventricular site (double arrow) with the three-dimensional optical flow (case #1). The three-dimensional optical flow of 1st and subsequent series (2nd ~ 5th) yields the displacement of selected points during the surgery.
Figure 4: Colored magnitude map of the displacement vectors overlayed on the image scanned before the craniotomy (case #1, FOV 240 mm). Cortical area of up to 15 mm deep shift up to 8 mm.

Influence of contrast agent and tumor removal

Figure 5 illustrates the influence of the contrast agent (Gadolinium-DTP Areg.) on the optical flow measurement. The contrast agent injected before dura opening enhances the tumor by highlighting the microvessels. The injection is necessary to visualize the lesion and its periphery for complete resection. The drawback of the contrast agent to the optical flow measurement is that the agent also highlights microvessels in the skin and thickens the skin signal in MRIs. Our method assumes that these intensity changes are due to motion, which causes some error in this situation.

We avoid this problem by segmenting the brain as a region of interest, thus eliminating artifacts arising from the skin intensity change. Results shown in experimental studies are all from these segmented grayscale images. The segmentation was accomplished by manual outlining assisted by basic image processing; morphological operations, island removal, and thresholding.
We didn’t observe the significant deterioration of the images in gray and white matter. In theory, intensity-based optical flow is vulnerable to image noise. Introduction of a more robust optical flow method, e.g. phase based optical flow, might be necessary when we observe more significant noises in the images.

Figure 5: Influence of contrast agent (gadolinium-DTPA) on optical flow measurement (Left: deformation analysis without removal of skin and tumor. Right: with removal of skin and tumor). Note that the analysis without removal of skin and tumor (right) shows unreasonable skin deformation in the lateral side, which cannot be observed when skin and tumor are removed before measurement (left). The agent for tumor enhancement also highlights microvessels in skin and thickens the skin signal in MRI.
Discussion

The method described in this paper is based on Horn and Schunck’s two-dimensional optical flow measurements. Our contribution was to expand the two-dimensional optical flow to three-dimensional employing a similar smoothness term in the optimization: the sum of squares of partial derivatives of displacements with respect to coordinate axes. This model applies when we can assume that neighboring displacements are similar, though some studies question this assumption in real camera imagery and propose other methods with different smoothness constraints. Alternatively, we could employ a smoothness model based on the assumption that the brain deformation field is mostly continuous with occasional discontinuities.

The accuracy (average error 3.5mm) of the method from the phantom test suggests that the method may be applicable to the study of brain deformation in surgeries; yet further refinement of the method is necessary to perform surgical navigation guided by these updated images.

Trobaugh et al. (12) investigated intraoperative brain shift by mounting an ultrasound probe on a tracking sensor and coregistering preoperative MR/CT with intraoperative ultrasound images. The registered ultrasound images were helpful in predicting intraoperative brain deformation and updating mismatched preoperative images. Another publication (13) discusses three categories of depth movements. Hill (2) and Roberts (4) were among the first groups to tackle measurement and quantitative analysis of intraoperative brain surface shift. Hill reported a study measuring the deformation of the dura and brain surfaces between the time of imaging and the time right after dura opening and before resection. In the surgery, the patients were first registered to
preoperative images by localization of fiducial markers and standard patient-image registration in an image-guided surgery system. Then, they measured the location of multiple points (approx. 60) on the brain surface to compare their position with respect to those in the preoperative images. They reported mean displacements of the brain surface of 5.6 mm before tumor resection. Roberts’s study used a ceiling-mounted robot microscope system to measure surface displacement during surgery. They determined mean displacement of approximately 10 mm and a presumed gravity direction of movement.

Freeborough and Fox, (14) measured brain deformation to track Alzheimer’s disease by nonlinear registration modified from (15). Maurer et al. measured the brain shift from pre- and postoperative images scanned in interventional MR setting (3) with voxel-similarity rigid registration described by Studholme et al. (16). No qualitative findings about subsurface displacement were in their report, except a preliminary result from three-dimensional B-spline-driven deformable registration.

Both the phatom and the clinical test indicated that the proposed method is capable of capturing surface and subsurface shift of the brain. It also enables the deformable registration of predeformation images to postdeformation images by applying the computed motion field to the predeformation images. Such functionality provides the means to warp intraoperative images to match preoperative images with fine anatomical and physiological details. We are also interested in the possibility of predicting the deformation in advance by extrapolating the computed motion fields in sequential intraoperative MRIs.
Conclusions

We developed a three-dimensional optical flow method to measure volumetric brain deformation from sequential intraoperative MRIs by expanding an existing gradient-based two-dimensional optical flow. A phantom study by a deformable gel showed that the method has an accuracy of 3.5mm. The clinical results from five cases exhibited that the method is capable of capturing surface and subsurface shift of the brain.
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