Fusion of coregistered cross-modality images using a temporally alternating display method

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Abstract—*As an aid to the interpretation of functional images, cross-modality coregistration of functional and anatomical images has grown rapidly. Various ways of easily interpreting and visualising coregistered images have previously been investigated; for their display, an intensity-weighted temporally alternating method is used. For brain images, geometric registration involves the automatic alignment method, using the head scalp boundary extracted from the sinogram of a PET emission scan and a surface-matching algorithm; images of the chest or abdomen are registered semi-automatically using a paired point matching algorithm. For the simultaneous display of geometrically registered images, rapid image switching is applied; both images are written with independent colour scales. The rapidly alternating display of two images, synchronised with monitor scanning, induces the fusion of images in the human visual perception system. The accuracy of registration of PET and MRI images is within 2 mm for two point sets. A resulting image is intensified by weighting the display time and/or controlling the intensity map of each image with the degree of interest. This method may be useful for the interpretation and visualisation of coregistered images.*

Keywords—*Coregistration, Visualisation, Fusion, PET*

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1 Introduction

CROSS-MODALITY COREGISTRATION of functional, positron emission tomography (PET) or single photon emission computed tomography (SPECT) and anatomical images, magnetic resonance imaging (MRI) or computed tomography (CT) can enhance and co-ordinate clinical information from both modalities (TAYLOR *et al.*, 1996). In recent years, applications of this technique have been widely used in the areas of diagnosis, pre-operative evaluation, image-guided surgery, functional brain mapping and the planning of radiotherapy (WAHL *et al.*, 1993; GRIMSON *et al.*, 1996; PETERS *et al.*, 1996; ZUBAL *et al.*, 1995; MOLINA *et al.*, 1995; MELTZER *et al.*, 1996).

In clinical studies of the human brain, the coregistration of brain PET and MRI has become increasingly useful for the localisation of lesions, such as the hypermetabolic epileptogenic zone and brain tumours. To improve the performance of registration and the effectiveness of the visualisation of coregistered images, several methods have been proposed (REHM *et al.*, 1994; PELIZZARI *et al.*, 1989; LEVIN *et al.*, 1989; BESL and MCKAY, 1992; MEYER *et al.*, 1992; PIETRZYK *et al.*, 1994).

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Visualisation has been an important technical issue. Various visualisation techniques have thus been investigated so that the coregistered images can be easily interpreted (WAHL *et al.*, 1993; REHM *et al.*, 1994). These include pixel interleaving and alternate interlacing of the even and odd fields of a video monitor, which have been widely used. The disadvantages of these methods, however, are a loss of spatial resolution and a lack of imaging flexibility.

In this study, we propose a new method for the simultaneous display of coregistered CT/MR and PET images. The basic idea is simple: a rapidly alternating display of both images with independent colour scales allows the fusion of images in the human visual perception system. In addition, a particular aspect of the fused image can be emphasised by increasing the display time and/or the intensity of one image with respect to the other. Advantages of this method are that, in comparison with the widely used pixel-interleaving method, the spatial resolution of the resulting image is not lost, and its intensity can be easily modified to emphasise functional or anatomical aspects.

2 Method

2.1 *Data sets*

PET: Doses of 3.7×10^8 Bq (10 mCi) of 2-[^{18}F]-fluoro-2-deoxy-D-glucose (FDG) were injected, and PET scanning was carried out on the ECAT EXACT 47* scanner. Before

*Siemens-CTI, Knoxville, USA

FDG administration, transmission scanning was performed using a Ge-68 rod source for attenuation correction. The location of the cross-hairs from a laser alignment system was marked on the patient, who was then removed from the scanner and injected with 370 MBq (10 mCi) of FDG. Static emission scans were started 30 min after the injection and continued for 30 min. Trans-axial images were reconstructed with a Shepp filter (cutoff = 0.30 cycles pixel^{-1}) as $128 \times 128 \times 47$ matrices with a pixel size of $2.1 \times 2.1 \times 3.4$ mm.

MRI: Brain MRI studies were performed on a 1.5 T magnet system†. T1 weighted MRI data were stored as 256×256 matrices with a size of 0.82×0.82 mm. The slice number was 20, and the thickness of the slices and the distance between them were 1.5 and 5.0 mm, respectively.

CT: For CT scanning, a GE SOMA+† was used. Images were sampled as 512×512 matrices with a trans-axial thickness of 10 mm.

2.2 *Geometric registration*

A program for three-dimensional coregistration was developed using a MATLAB 4.2c‡, and, for geometric registration (scaling, translation and rotation) of the images, two methods were used. Brain MRI and PET images were registered using a three-dimensional surface-matching technique (LEE *et al.*, 1998). The surfaces of the head scalp were extracted from MRI and PET images. Boundary extraction of the head from a PET image is one of the most important technical problems of registration. Using the sinogram of a PET emission scan, we tried to ensure that the extraction of the head boundary was robust, and we used a boundary-enhanced PET image to coregister the PET and MRI modalities. Pixels with 10% of the global maximum pixel value were considered to be boundary pixels of the sinogram. The values of these boundary pixels were replaced with the maximum pixel value of the sinogram to enhance its boundary. Using the boundary-enhanced sinogram, we reconstructed trans-axial slices, from which the boundary of the head scalp was extracted. As displacement errors due to patient repositioning between transmission and emission scans were eliminated and it was not necessary to remove the couch, this method was superior to boundary extraction using transmission images.

Using a simple threshold method, 180 boundary points were extracted at intervals of about 2° from each slice of the MR images. Best affine transformation between the two point sets was performed using the least-square fitting, which minimised the sum of the Euclidean distance between the point sets (TAYLOR *et al.*, 1996; PELIZZARI *et al.*, 1989; LEVIN *et al.*, 1989). Using a pre-defined distance map, we reduced the calculation time (TAYLOR *et al.*, 1996; BORGEFORS, 1988; ITTI *et al.*, 1997).

Abdomen/chest CT and PET images were coregistered by means of a paired point matching algorithm that uses user-identified paired points and finds the affine solution by minimising the root-mean-square distance between all paired points (BESL and MCKAY, 1992; MEYER *et al.*, 1992).

All the images were saved in a 256×256 matrix format after geometric registration.

2.3 *Display of coregistered images*

For the simultaneous display of the geometrically registered images, we used a video card** and 17-inch high resolution monitor††. In the video card, a video BIOS supports a VESA BIOS EXTENSION (VBE) interface. In this study, we used the VESA VBE functions to initialise the graphics mode and switch rapidly between the two different displayed screens. The VBE standard defines a set of extensions to the VGA ROM BIOS services that can be accessed under DOS through interrupt 10 h‡‡. We initialised the VGA mode to be 1280×1024 at 87 Hz, 256 colour graphic mode through VBE function 02 h (set VBE mode) by placing the mode number (107 h for the resolution of 1280×1024 and 256 colours) in BL and clearing the upper byte (BH).

After initialisation, both 256×256 matrix images (PET and CT/MR) were loaded into the video memory with independent colour scales (PET: hot-metal; CT/MR: grey). The pixel in the upper left corner of each image was allocated to (384, 256) and (768, 256), respectively, and (0, 0) was selected as the pixel to be displayed in the upper left corner of the monitor. The PET image was thus displayed in the centre and the CT/MR image was displayed on the right side of the display. By shifting the display window to the left by the distance between the two images (384 pixels), the positions of the PET and CT/MR images could be changed from (384, 256) and (768, 256) to (0, 256) and (384, 256) on the absolute co-ordinate of the display. The display window was then shifted to the right, and PET and CT/MR images were thus alternately displayed at (384, 256). To induce optical fusion such that both images were overlaid, the method above was applied to rapid switching of the images. The procedure is demonstrated graphically in Fig. 1. The images were switched frame by frame; switching was synchronised with the monitor scanning through VBE function 07 h (set/get display start) by placing 80 h (set display start during vertical retrace) in BL and clearing the upper byte (BH). There was no interval between the display of the two images. The temporal modulation rate for the images was 87 Hz, as we synchronised the display of images with the vertical scanning.

The number of sequential frames, which were synchronised with the vertical scanning, was designed to be controlled according to the degree of intensification (i.e. PET-PET-MRI-P-P-M-..., M-M-P-M-M-P-...). By applying this time-weighted display alternation, we could easily control the intensification of the emphasis on an interesting image. The successive display of one image at more than three times the monitor scanning frequency gave rise, however, to a flickering effect. This phenomenon can be explained by the characteristics of the human visual system, which perceives flickering if the frequency is less than a certain critical level. To solve this problem, we designed an intensity-weighted alternating method, in which the numbers of sequential frames were equalised to be one for each modality, and the global intensity of each image was controlled with variable colour scales.

The resulting images were photographed at an exposure of 2 s duration while they were displayed on the monitor screen.

3 Results

For geometric registration of brain MRI and PET, the accuracy of our method was within 2.0 mm, as determined by the Euclidean distance between the two point sets extracted previously (LEE *et al.*, 1998). By qualitative visual assessment, successful coregistration of both images was achieved in abdomen/chest CT and PET images.

Figs 2 and 3 show the results of the time-weighted and intensity-weighted alternating methods, respectively. The images in the upper rows are the results of the brain PET and

†GE, Milwaukee, USA

‡MathWorks, Natick, USA

**3D RAGE II+, ATI Technologies Inc., Thornhill, ON, Canada

††HiSync DT17XR, LG Electronics Inc., Seoul, Korea

‡‡Available at: http://www.vesa.org

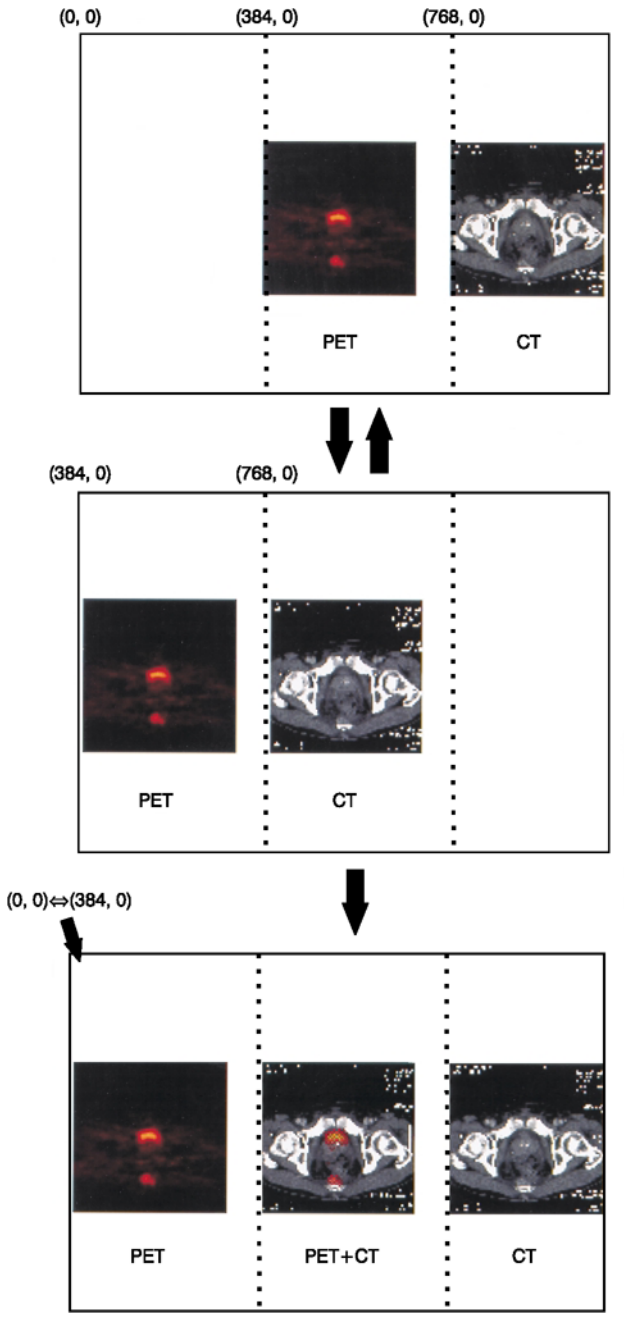

Fig. 1 *Schematic diagram of alternating method for display of geometrically coregistered images*

MRI registration, and those in the lower rows are those of the abdomen PET and CT registration.

In Fig. 2, the first and last images in each row are the geometrically coregistered PET and MRI/CT images, respectively. Simultaneously displayed images using the time-weighted alternating method are represented between them. The ratios in the middle represent the numbers of sequential frames (PET: MRI/CT). When we equalised the numbers of sequential frames (P-M-P-M- . . .), a similar effect seen in the pixel interleaving method was observed (third column). By increasing the sequential display time of one image, we could enhance its influence, such as in the second and fourth columns. However, this resulted in a flickering effect in latter cases.

Fig. 3 shows the simultaneously displayed images using the intensity-weighted alternating method, where the numbers of sequential frames were fixed at one and one, and the global intensities of each image were controlled using variable colour scales. The values in the middle represent the ratio of the intensities (PET: MRI/CT). This method could be used to generate the same images as shown in Fig. 2 without the flickering problem.

The images fused using these timely alternating display methods facilitated the accurate localisation of FDG uptake in the PET images. The resulting images of the brain PET and MRI registration in Figs 2 and 3 clearly demonstrate a normal FDG uptake in such MRI-defined structures as the nucleus caudatus, thalamus and both temporal lobes. Those of the abdomen PET and CT registration show that increased FDG uptake was well localised in the bladder and the tumour at the post-rectal pouch.

4 Discussion

As a tool for combining information about anatomy and function and improving localising ability, multimodality coregistration of medical images is widely used in many applications such as diagnosis, functional mapping and treatment planning. Although several methods for the visualisation of the resulting coregistered medical images have been investigated, most previous research has focused on the problem of geometric matching (REHM *et al.*, 1994; PELIZZARI *et al.*, 1989; LEVIN *et al.*, 1989; BESL and MCKAY, 1992; MEYER *et al.*, 1992; PIETRZYK *et al.*, 1994). The reason may be because improvement in the accuracy and processing time of geometric matching has so far been considered more important. For the effective application of coregistration to the clinical situation, it is necessary and important, however, to visualise the resulting co-registered images.

Common approaches to the problem have included superimposing the contours extracted from the high-resolution onto the low-resolution image, and interleaving alternate pixels with independent colour scales. The superimposing tactic has certain disadvantages: the accurate segmentation of each anatomical structure, in itself, requires that the processing technique be at a very high level. In addition, pathologically induced anatomical variations and structural changes make the process more difficult and decrease the robustness and reliability of segmentation. Even if boundaries can be extracted exactly and successfully overlaid on images with lower resolution, the use of partial information from higher-resolution images, such as CT and MRI, is not effective.

To overcome this problem, the alternate pixel interleaving method has been suggested; this makes equal use of information from both modalities and gives the appearance of one partially transparent volume overlaying another (REHM *et al.*, 1994). In this approach, each pixel alternately displayed on the screen is derived from each of two images, without any algebraic manipulation. Individual images can be presented with different colour scales that can be managed in a parallel way. The loss of spatial resolution can, however, cause certain problems: as this method depends only on perceptual fusion with adjacent pixels, this camouflage may fail to induce fusion, especially in areas with high contrast between both images.

In this study, we propose a new strategy where coregistered images can be simultaneously visualised. Rapid alternation of the screen on which each image is displayed with independent colour scales causes the fusion of both images but no loss of spatial resolution.

Using Bloch's law, we can explain how the visual system accomplishes this temporal summation or fusion of two images (HALLETT, 1991). The effect of a brief impulse input of photons decays with time t as $\exp(-t/\tau)$. In the case of human rod vision, $\tau \approx 0.1$ s over most conditions (this time constant is much longer than that of the exponential decay of display phosphor luminescence; the effect of persistence of the phosphor can therefore be ignored). In the absence of a background, complete temporal

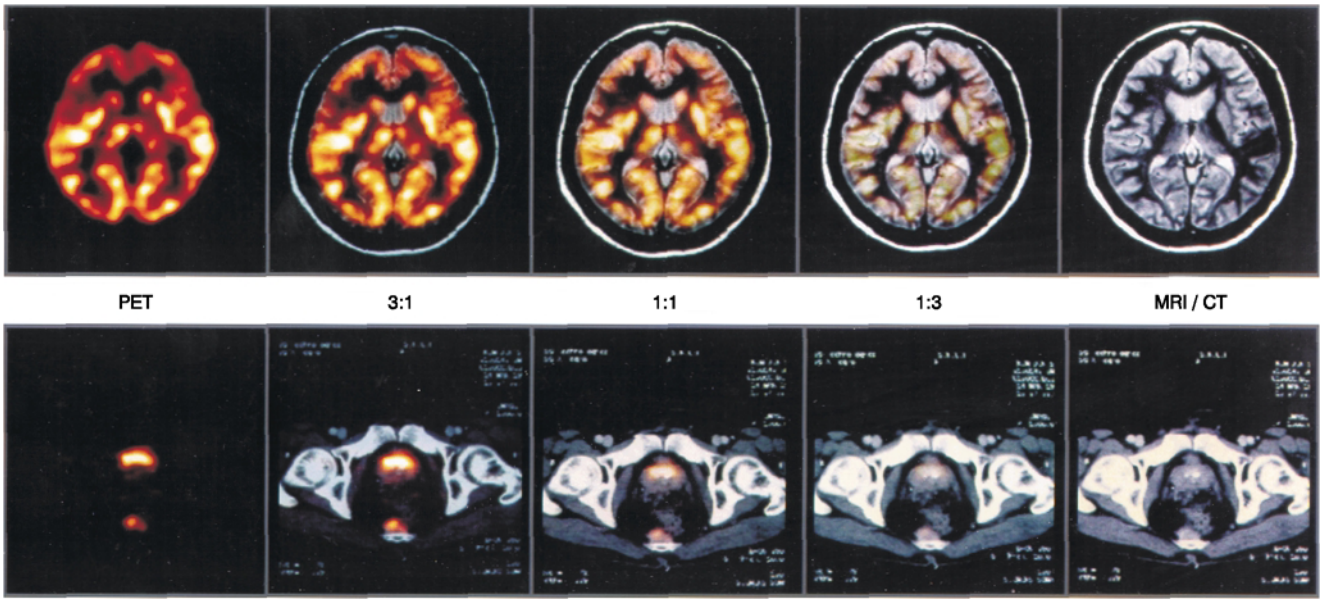

Fig. 2 *Time-weighted alternating method: numbers of sequential frames, synchronised with vertical scanning, were controlled according to degree of intensification. Ratios on centreline represent numbers of sequential frames. Images were photographed at exposure of 2 s duration*

integration of energy for a just visible stimulus extends to $\tau \approx 0.1$ s, independent of the test area (HALLETT, 1991). That is, if a visual stimulus occurs before the delayed effect of the previous one has disappeared, the sequential stimuli will overlap to form a fusion of their effects. Temporal sensitivity of the human visual system therefore has a finite limitation. However, it is hard to define the frequency threshold above which the eye does not discern the flickering effect as temporal sensitivity of the human visual system is dependent on the mean background intensity. In this system, the highest detectable temporal frequency ranges from 30–40 Hz at low ambient light levels to 80 Hz at high ambient light levels (WANDELL, 1995; WATSON *et al.*, 1986; FARREL, 1991). Fortunately, computer monitors currently in use in the clinical environment scan at a sufficiently high frequency. The problem of flickering does not therefore exist when images are temporally alternated.

When displaying two images, it is necessary to emphasise the information provided by one without entirely suppressing that provided by the other. When applying co-registration as the supporting tool for interpreting SPECT or PET, some degree of structural information from CT or MRI should appear, so as not to obscure the functional information provided by SPECT or PET. By increasing the display time of the selected image, the more interesting one can be intensified, and the effect of the lesser one will be reduced. When the display time of one image is over-extended, however, this time-weighted alternating method begins to suffer from the problem of the flickering effect. The intensity-weighted temporally alternating method was thus designed to overcome this flickering problem in which the overall intensity of each image was controlled while the ratio of the displaying period was equalised. Furthermore, giving an observer the ability to change the

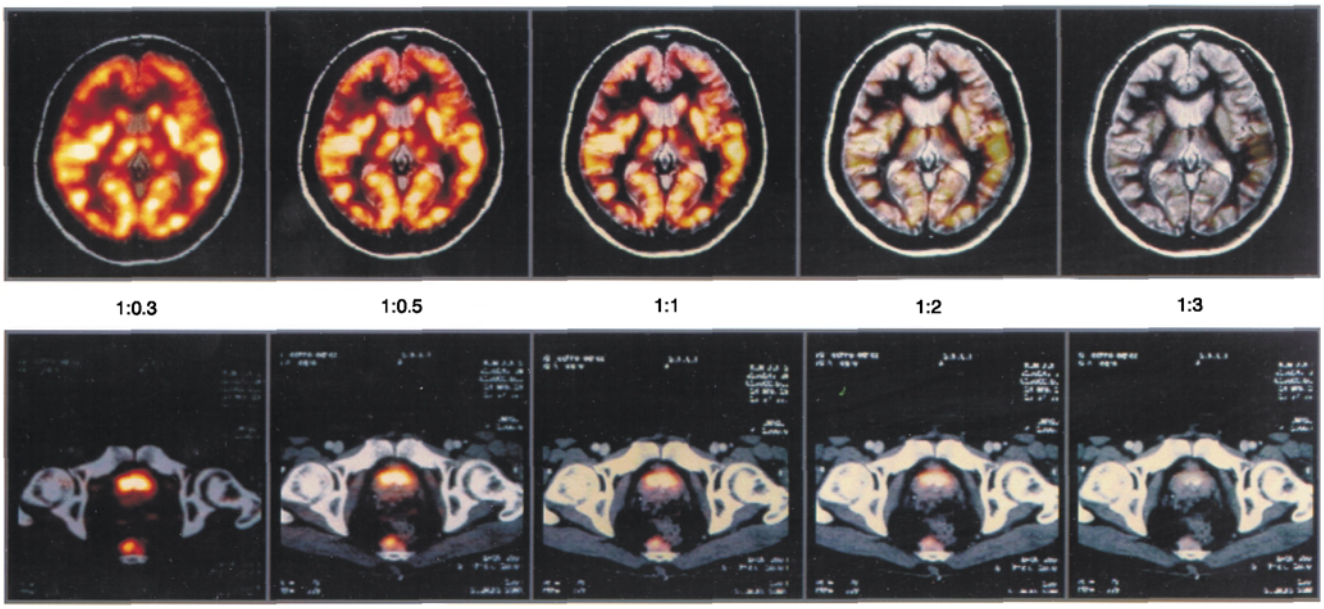

Fig. 3 *Intensity-weighted alternating method: numbers of sequential frames were fixed at one and one, and global intensities of each image were controlled with variable colour scales. By applying this method, problem of flickering was solved. Values on centreline represent ratio of intensities (PET: MRI/CT). Images were photographed at exposure of 2 s duration*

relative intensity of the two images interactively could prevent the problem of potential loss of information when the two intensity scales are added.

In comparison with the analytic composing for the static combination of images which models the human visual system to render one or both of the images semi-transparently, our approach has a unique advantage, even though the analytic method may be able to provide a similar effect to ours. To generate exactly the same effect as an alternating display, too many kinds of colour need to be provided in analytic composing. If we let N be the length of a colourmap for an image and assume that each image uses a different colourmap, the number of colours for analytic composition will be proportional to N^2, although the alternating display method needs only $2N$ colours. The reason is that, in the worst case in the analytic method, every pair of colours from each colourmap will generate a new composite colour. Therefore we can say that our method has the benefit of saving computation time or load needed to generate an analytic composition. On the other hand, an important issue requiring further study in our approach is the difficulty of generating a hard-copy image that precisely matches the image composed by the alternating display. Proper mathematical modelling and analytic composition will provide the solution.

The intensity-weighted alternating display method also required the implementation of an analytic process to generate a colourmap whenever the user wanted to change the relative intensity of two images. We implemented this procedure by generating a new colourmap through simple multiplication of a given constant with the original colourmap. This procedure is so trivial and elementary that it can be completed within a very short period of time and the user should not feel any dissatisfaction.

In this study, we used an 8-bit display (1280×1024 at 87 Hz; 256 colour graphic mode) for the grey-scale CT and MR images. This would seem to limit the clinical usage of our methods to situations in which the MR or CT image was being used solely as a guide to anatomical boundaries. However, a 16-bit display of images could be accomplished easily if we used 1024×768, 16-bit graphic mode with 4 MB video RAM. If we wish to maintain the resolution of 1280×1024 with a 16-bit graphic mode, we can just add some more video RAM to the video card.

More investigations are needed to justify our temporally alternating method. In particular, more physical and/or psychophysical experiments to measure spatial resolution more objectively are needed to reach a firm conclusion about the advantage in terms of the spatial resolution of our method.

5 Conclusions

We have proposed a new method for the simultaneous display of coregistered functional and anatomical images. A rapidly alternating display, synchronised with monitor scanning, induces fusion in the visual perception system such that both images are overlaid. After coregistration, one modality image can be intensified with respect to the other by a time/intensity-weighted alternating display; the intensity-weighted alternating method can solve the problem of flickering. In conclusion, the intensity-weighted temporally alternating method is a practical solution that allows for the fusion of cross-modality medical images. These techniques are useful for the interpretation and visualisation of co-registered images.

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