Radiofrequency ablation (RFA) as a source of destructive energy is currently the most studied and widespread method of local tissue destruction, and is also implemented in numerous commercially available equipment. Despite certain limitations, RFA remains the standard against which other types of percutaneous energy exposure are usually compared in terms of effectiveness.

Thermal ablation techniques have many similarities, and certain developments and technical solutions related to the use of medical imaging techniques to monitor, for example, microwave ablation (MCA) can be extrapolated to some extent to RFA and vice versa.

The medical imaging method in terms of providing ablation technologies (AT) has three separate but related tasks ensuring the optimal location of the electrode (antenna, probe, light guide, applicator, etc.) for the procedure; providing real–time monitoring of the spread of the ablation zone; and assessing the effectiveness of the ablation performed [1]. Since these tasks require the acquisition of multi–plane images and their three–dimensional reconstructions, there are not many suitable methods, such as ultrasound (US), computed tomography (CT) and magnetic resonance imaging (MRI) [2]. However, none of these methods can solve all three tasks with equal success, because each of them has its own advantages and disadvantages, and each task, since it has its own rating of requirements, is currently being solved independently of other tasks, and therefore, attempts are still being made to either improve diagnostic methods or combine the advantages of different methods in one technological solution.

Control over the percutaneous advancement of the applicator to the tumour site requires adequate real–time visualisation of the probe and target. In the vast majority of observations, adequate visualisation quality can be achieved using any of these methods [3, 4], but only ultrasound can provide real–time control [2]. Additional advantages of this method are portability, the possibility of using intracavitary sensors, low cost and wide availability; moreover, in radiology, ultrasound is considered a theranostic technology for cancer, i.e. a method that combines therapeutic and diagnostic qualities [5]. However, the quality of imaging is the worst, as the resolution of ultrasound is determined by the ultrasound wavelength, i.e. the operating frequency of the transducer. It can be increased by using high–frequency transducers, but at a short distance, since the depth of penetration of the ultrasound wave is inversely proportional to its frequency; therefore, to examine the liver of an adult at its full depth, transducers with a frequency of approximately 3.5 MHz should be used. In addition, due to the well–known falsity of the postulates about the propagation of ultrasound waves in the whole body, artefacts may be present in the form of false images of non–existent objects or the absence of images of real structures, because to obtain an image of an object, it is necessary, firstly, that its acoustic density differs from the density of the surrounding tissue, and secondly, that the wave reflection has sufficient energy to return to the transducer, and ultrasound waves attenuate with distance. That is why, for example, up to 25% of small hepatocellular carcinomas (HCC) in the setting of liver cirrhosis are not visualised by ultrasound [6]. It should also be borne in mind that this technique is highly dependent on the operator's experience, and certain areas of the liver during ultrasound are so–called blind spots, which are extremely difficult to visualise (liver dome, lateral edge of the left lobe and areas adjacent to the ribs), and the ultrasound window may be additionally limited by the adjacent colon or omentum [2, 7]. Given these disadvantages, it is surprising that most RFAs in the world, especially in Asian countries, are still performed under ultrasound guidance. This may be explained by the widespread use in this region of Sonazoid, a contrast agent based on microspheres with perfluorobutane developed in 2007 (Daiichi Sankyo, Tokyo, Japan). In contrast to Levovist® (Shering, Berlin, Germany), SonoVie® (Bracco Imaging, Milan, Italy) contrast agents, which enhance blood vessels directly and have a rather short time window (the most informative arterial phase lasts from about 30 to 50 seconds after injection), Sonazoid also actively phagocytes cells of the reticuloendothelial system and has an additional "Kupffer" phase of contrast (approximately 10 minutes after injection), on which liver tumours or liver metastases appear as accumulation defects against the background of enhanced surrounding parenchyma, as the number of Kupffer cells in tumours is much lower, which simplifies the positioning of the applicator [8].

In contrast to sonography, liver MRI usually provides the highest image quality and clearly demonstrates the boundary between the tumour and unchanged tissue [9, 10]. However, obtaining a magnetic resonance image requires maximum time and the availability of not only the magnetic resonance imager itself (or two – 1.5 T for planning and 0.2–0.4 T for treatment), but also special amplified ablation equipment capable of operating in a powerful magnetic field [11], non–
standard MRI and a special set of sequences [12], which is why MRI–guided RFA remains the property of a narrow range of specialised clinics.

CT control of RFA applicator insertion occupies an intermediate position between ultrasound and MRI in terms of imaging quality, image acquisition time, equipment availability, procedure complexity, and cost [13, 14]. Unlike ultrasound, CT provides a three–dimensional image of the tumour and surrounding structures, and intravenous contrast facilitates visualisation of blood vessels and allows for assessment of tissue vascularisation. Its use is necessary for the ablation of tumours that are not visible during ultrasound or located in the liver dome, which requires the use of percutaneous access [15]. The disadvantages of CT include the presence of ionising radiation and limitations in the placement of the ablation transducer in the hepatic plane. Interestingly, some consider even the prolonged blocking of the CT device for diagnostic procedures during ablation to be a limitation of CT, promoting the concept of using hybrid operating rooms for AT, where one gantry sliding on rails moves, if necessary, between two tables that can be separated by a lead bulkhead [16]. However, the main limitation of CT for use in AT is the inability to obtain images in real time.

This disadvantage is eliminated by a modern type of CT – cone beam CT (CBCT), implemented on the basis of flat–detector angiographic systems and improved by image reconstruction algorithms. When the X–ray generator and image detector rotate around the patient’s body, axial CT–like images are obtained and their three–dimensional reconstruction is performed [17]. Moreover, the radiation dose is significantly lower than that of a conventional CT scan, as a much smaller volume is irradiated. Since the same C–arc is used to obtain the fluoroscopic image as for CBCT, it is possible to overlay the CBCT image on the fluoroscopy image in real time; this allows for tumour detection, semi–automatic segmentation, AT planning and control [18], including the use of navigation software for needle path planning, marking ablation zones and image fusion [19]. X. S. Yao and colleagues [18] reported complete visual control of RFA at virtually all stages in 48 patients with HCC. CBCT was performed before each RFA to plan the needle path and ablation zone, as well as after it to immediately assess the response to treatment. During percutaneous puncture and needle insertion, a combination of CT scan and real–time fluoroscopy images was used. The results of RFA coincided with the planning data with an accuracy of 91.7% (according to the mRECIST criteria, it was planned to achieve a complete response to the procedure in 87.5% and a partial response in 12.5% of patients, according to contrast–enhanced CT or MRI a week after RFA, a complete response was achieved in 79.2% and a partial response in 20.8% of patients) [20]. It can only be added that the study used one of the most expensive angiographic systems – Allura Xper FD20 with XperCT and XperGuide software by Philips Healthcare, the cost of which is not significantly different from the cost of the best magnetic resonance imaging systems.

Back in the early 90s, the advantages, or rather disadvantages, of individual methods stimulated the rethinking of medical imaging and the creation of new hybrid technologies under the general name “Fusion Imaging”, which were originally intended to combine images of the organ structure and its functioning [21]. The first prototype, created in the early 90s at the University of California under the leadership of B. H. Hasegawa in the form of a combination of CT and single–photon emission CT (SPECT), was introduced by General Electrics (GE) under the name “Hawkeye” in 1999, even before the commercial introduction of positron emission tomography (PET)/CT the following year [22]. The SPECT/CT hybrid was followed in 2001 by GE’s more successful and most widely used PET/CT combination in the world today, where PET was combined with CT imaging – Discovery LS, which was conceived in Geneva in 1991 by engineers D. Townsend and R. Nurr and oncological surgeon R. Egeli, who proposed to fill the gaps between the rotating blocks of bismuth and germanium oxide detectors on the PET scanner model with something useful, such as a computed tomography scanner, which could provide anatomical information more understandable for surgeons” [23].

For the further development of AT, it was of great importance to combine the diagnostic accuracy and orthogonality of CT or MRI with ultrasound by merging their images in real time in any arbitrary planes (in fact, the software of the ultrasound scanner software, together with the real–time ultrasound image, should generate a data set from the previously obtained CT or MRI image and provide a reformatted corresponding CT or MRI image on the second half of the screen or “underlay” the sonogram) [24]. To do this, it is necessary to ensure image scaling due to the coincidence of landmarks and tracking changes in the position of the ultrasound transducer in the CT or MRI coordinate system. Both optical tracking based on image analysis, which is most often used for surgical procedures, and electromagnetic tracking (EMT), which is most often used for ultrasound–guided interventions, are available for Fusion Imaging technology [25]. To implement EMT–based image fusion, at least three components are added to the ultrasound scanner: a magnetic field generator (external device), a position sensor (placed on the transducer handle), and a position sensor unit (located in the ultrasound scanner). The magnetic field generator, located next to the patient, generates a magnetic field, thereby inducing an electric current in the position sensor mounted on the ultrasound transducer. During the movements of the ultrasound transducer, the magnitude of the electric current of the position sensor changes with respect to the magnetic field; these changes are detected by the position sensor unit, which calculates the exact position of the position sensor and therefore the location, direction and magnitude of the movements of the ultrasound transducer [26].

Image fusion with EMV can be performed using external reference markers and/or internal anatomical landmarks. First, a CT or MRI scan is performed with external reference markers attached to the body surface around the liver, and the CT/MRI
data is uploaded to the ultrasound system. The ultrasound transducer then fixes their location in space by touching the position sensor, after which the images are automatically merged to create a plane. Often, internal markers, which are anatomical landmarks in the patient’s liver, such as branches of the hepatic or portal hepatic veins, cysts, calcifications, etc., are also combined. The coordinates of each marker are measured and the corresponding transformation matrix is calculated. If the markers are anatomical, such as blood vessels or specific points in organs, the images must have a high spatial resolution to ensure that the same point is marked in each image. Almost all image fusion techniques are based on the concept of a “solid”, i.e., on the assumptions that, first, the relative spatial relationships between all structures in the data set are identical in both images; second, the motions are affine transformations (i.e., straight lines and distance relations between elements are preserved). In reality, the data changes as tissue deformation occurs due to breathing, patient movement, or mismatching of the patient’s position during the examination, which can cause distortion that affects the entire image. To account for tissue deformation, a much more complex non-affine transformation, sometimes called “image warping,” is required, which must be based on a theoretical physical model of the tissue that describes its flexibility, stiffness, and tensile strength. This approach, called “finite element modelling”, is widely used in engineering where the physical properties of materials are well documented, but it is extremely limited in medical applications due to the extreme complexity of the body and the variability of tissue properties [26]. To date, quite a few methods for calculating automatic joint registration have been proposed, the review of which is more a subject of computer science and computer engineering than medicine, which is primarily interested in the accuracy of fusion, the time for fusion and the implementation of the technology in serial equipment [26–29].

The first serial production of ultrasound sensors and instruments for EMF biopsy was carried out in 1996 in Israel in the Ultraguide 1000 ultrasound system manufactured by the company of the same name [30], but it was not in demand and the company was forced to leave the market in 2003. More successful was the fate of similar products – Virtual Navigator (manufactured by Esaote S.p.a., Genoa, Italy) and Real-time Virtual Sonography (manufactured by Hitachi Medical Systems, Tokyo, Japan), which also used EMF as both an ultrasound transducer and an electrode for RFA, merging CT and MRI images with ultrasound images [31, 32]. The development of PercuNav by Traxtal Inc. (Toronto, Canada) in 2006 was implemented on the basis of the ultrasound diagnostic platform EPIQ7 by Philips Healthcare (Best, the Netherlands) and used miniature electromagnetic tracking sensors embedded in the tips of ablation and biopsy needles, providing accurate navigation even for flexible instruments. Well-known systems such as the American Volume Navigation, German eSie, Korean S-Fusion, Japanese Smart Fusion and Smart Navi are implemented on the ultrasound platforms LOGIQ E9 (manufactured by GE Healthcare, Milwaukee, USA), ACUSON S3000 (Siemens, Erlangen, Germany), RS80A (Samsung Medison, Seoul, Korea), Aplio Platinum (Canon Medical Systems, Tochigi, Japan), some of them also for precision medical imaging: CIVCO Medical Solutions (Coralville, USA) or eSi Guide Needle Tracking (Siemens, Erlangen, Germany) [33–35].

The spatial accuracy of imaging was evaluated using these systems in many experimental and clinical studies [26, 31, 36–39]. The average tracking error was about 8 mm, reaching the highest values in the dome of the liver and the lateral part of its left lobe, and the maximum accuracy of (1.9 ± 1.4) mm was achieved when performing ultrasound immediately after CT under general anaesthesia [26].

The main limitation of these systems is the lack of compensation for patient breathing and movement, which causes data mismatch. To optimise image fusion, co-recording should be performed in the same breathing phase as the previously acquired dataset, and the patient should be in the same position as during CT or MRI (i.e., in the breath-hold phase), especially when studying organs that actively move during breathing, such as the liver [2, 26, 33]. It is possible to compensate for the patient’s respiratory movements by placing external electromagnetic position sensors on the patient’s body, which is implemented in the Resona 7 ultrasound scanner on the ZST+ platform (joint development of the American company Zonar and the Chinese company Mindray) [40, 41]. Another disadvantage is the poorer visualisation of peripherally located small foci due to both their remoteseness from large hepatic vessels, which are most often used as internal markers for image fusion, and the more pronounced effect of respiratory movements on the liver periphery [42]. The use of contrast-enhanced ultrasound instead of standard gray-scale ultrasound should help to mitigate the risks [43].

Image fusion technology allowed not only to significantly increase the effectiveness of AT under sonographic control, but also to stimulate the development of CT as a control method by combining spatial resolution, three-dimensional coordinate system and frameless stereotactic CT navigation systems with real-time ultrasound observation called “stereotactic thermal ablation” [44, 45]. The greatest contribution to its development was made by the Department of Interventional Oncology at the University Hospital of Innsbruck (Austria), where RFA (2001), MHA (2008), irreversible electroporation (2013) and cryoablation (2014) were performed in a stereotactic way for the first time [46]. The hallmarks of this group of researchers are:

- the use of optical three-dimensional navigation systems (S8 manufactured by Medtronic Inc., USA, or CAS—ONE, CAScination AG, Switzerland) with planning of multiple trajectories on multi-plane reformatted sections;
- ATLAS sight (Interventional Systems Inc., Kitzbühel, Austria) and 15G/17.2 cm coaxial needles;
- superimposition of the CT image after needle placement on the primary “planning” CT scan;
- insertion of up to three electrodes for water-cooled radio frequency (RF) through coaxial needles;
control of the patient's respiratory movements (complete muscle relaxation during anaesthesia, high-frequency flow ventilation, or transnasal humidified rapid insufflation [44, 45]). This allows for the ablation of large (up to 8 cm in diameter) and multiple (up to 24) lesions in a single session [46] with an average error of (1.98 ± 0.93) mm (range 0.44 – 4.02 mm) [47] and targeting accuracy for 145 needle tips (3.6 ± 2.5 mm) [48]. Despite the clear advantages of MHA over RFA [49], these researchers remain principled supporters of the latter, motivated by the possibility of better modulation of the necrosis zone by smaller areas of thermal exposure, lower cost of electrodes for RFA, and larger antenna diameters (typically 13G for MHA and 17G for RFA), which makes it impossible to insert them through coaxial needles [46].

It should be noted that stereotactic thermoablation of hepatic metastases of colorectal cancer (CRC) with similar software for planning, navigation and image fusion is used in the world, in addition to the above—mentioned group, only in two centres – university clinics in Bern (Inselspital) and Stockholm (Danderyd Hospital, Karolinska Institutet). The main differences are the use of MHA, the non—use of coaxial needles, other ventilation techniques, and the use of sterile reflective markers to track patient movements. According to the first centre, in the treatment of 301 primary and secondary liver tumours during 191 ablation sessions, the average positioning error per probe was (2.9 ± 2.3) mm, but the average diameter of the foci was only 1.5 cm [50]; second, the average lateral error of the antennas, depth error, and total error were respectively (4.0 ± 2.5), (3.4 ± 3.2), (5.8 ± 3.2) mm with an average tumour diameter of (14.9 ± 5.9) mm [51, 52].

However, even with the ideal positioning of the destructive energy applicator, the ratio of tumour size to the ablation zone is crucial: to prevent under— or over—destruction, it is necessary to have reliable, et al. accurate and fast, control over the ablation process. Contrary to initial enthusiasm,grey—scale ultrasound data during hyperthermic ablation is not accurate enough to predict the degree of coagulation [53]. The hyperechogenic zone that occurs around the distal part of the applicator during energy application and is caused by microbubbles of gas that form in the heated tissue does not indicate coagulation. This area can vary in size, shape, and contour; it is progressively enlarged and can often conceal the energy applicator and the tumour [4]. Conventional colour and energy Doppler have also not proven useful in assessing the degree of induced coagulation, but ultrasound contrast immediately after ablation is used by many authors – if contrast enhancement is recorded at the tumour edge, another warm—up cycle is recommended [54, 55]. There have been attempts to use the overlay of application software packages to monitor the course of RFA [56], for example, RFA–Guardian [57]). Some have proposed methods of modelling the distant contour of the ablation zone, which is the worst visualised, using the so—called active contour model based on the leading edge contours of real—time and three—dimensional ultrasound images [58].

However, the new modalities of ultrasound equipment that have emerged in recent years, although they have not yet ushered in a new period of its flourishing as a method of controlling thermal ablation, give rise to serious hopes for this. The fact is that heating of biological tissue is accompanied not only by thermal expansion, but also by changes in a number of its mechanical (and hence acoustic) properties, which can be recorded during the processing of the reflected ultrasound signal – from the wave speed to the intensity of its scattering due to changes in the tissue structure caused by protein denaturation. There are known attempts to use the characteristics of the time and frequency components of the raw ultrasound backscattered signals of the heating area, including time and frequency shifts, attenuation coefficient, changes in backscattering energy, its probability distribution, changes in image texture in the B—mode, and tracking of spectra, which were performed over the entire temperature range during thermal ablation therapy on biological tissues both ex vivo and in vivo, and phantoms [5, 59–61]. To date, the following methods have been proposed for assessing thermal damage after thermal ablation: echo decoherence and integrated backscattering [62], determination of attenuation parameters [63], Nakagami parameter [64], determination of the average distance between scatterers [65], attenuation coefficient in A—mode sonography [61], assessment of time shift with the determination of the adaptive coefficient [66].

Attempts to estimate attenuation by numerous methods of analysing its time and frequency components have proved to be quite popular: envelope peak method, gain, spectral shift methods, spectral difference methods, attenuation coefficient, differential attenuation coefficient [67]. A large number of experimental works have been devoted to the difference in acoustic attenuation between thermally coagulated and untreated liver tissue due to heating or thermal ablation: an increase in the attenuation coefficient was demonstrated ex vivo in the liver of cattle, pigs, dogs, and rats [68]. For example, the attenuation coefficient in the areas of thermal damage generated in the excised porcine liver using an ultrasound therapy system increased from 3.72 to 7.22 dB/cm [69]. A sharp increase in the attenuation coefficient from 1.9 to 6.9 dB/cm occurred after thermal coagulation in bovine liver ex vivo [70], and in porcine liver in vivo during MCA it increased from 0.2 to 0.9 dB/cm [71]. However, despite these results and a rather long history, all these methods are still only experimental for monitoring liver RFA; in the clinic, some of them are used mainly to assess the degree of liver steatosis.

A relatively new method that is gaining more and more attention among researchers of percutaneous ablation of liver tumours is ultrasound elastography (UE), which provides a qualitative or quantitative measurement of tissue elasticity [72, 73]. The physical meaning of its use is that heat—induced cell necrosis leads to an increase in tissue stiffness and elasticity, and hence to an increase in Young's modulus and shear [74, 75]. Measurements of elastic properties on porcine liver tissue subjected to RFA in vivo show that parenchymal stiffness correlates quite closely with the ablation morphology,
as indicated by the colour of the tissue at autopsy – with a central "white zone" of coagulation necrosis surrounded by a transient "red zone" of hyperaemia (36.2 and 3.1 kPa, respectively) [76, 77]. To date, several types of UE have been proposed, and most of them have already been implemented in commercial medical equipment.

Rather unexpected ways of controlling the applicator position and monitoring RFA are also known. Photoacoustic imaging has been proposed as a potential complement to ultrasound during ablation: the positioning of the applicator is controlled by its location inside an annular fibre–optic probe, and the use of interstitial illumination is justified for differentiating between coagulated and native liver tissue based on ex vivo studies [78].

Native CT of the liver performed immediately after ablation usually shows increased density in the centre of the treatment area, often surrounded by a hypodense zone, except for encapsulated lesions (such as HCC), the boundaries of this outer hypodense zone are too blurred to provide sufficient sensitivity to evaluate therapy [79]. However, contrast–enhanced CT can also detect remnants of viable tumour immediately after thermal ablation, as it should demonstrate a hypodense zone at the ablation site devoid of characteristic tumour or parenchymal enhancement. In intrahepatic metastases, differentiation of coagulation necrosis from hypodense tumour is usually easiest on images in the equilibrium phase of contrast enhancement (5 to 10 min after iodine–containing contrast injection). During this phase, persistent hypodense is seen in coagulated tissues but not in viable tumour. Arterial phase images are most useful in HCC with early enhancement. Arterial phase imaging can also reveal a contrast–enhanced halo rim, which corresponds to an early inflammatory response to thermal injury; this inflammatory rim can be seen immediately after ablation and regresses within the first month after treatment [80].

In addition to the unsurpassed quality of tissue visualization, which provides accurate positioning of the applicator in the focus on a three–dimensional image, a key advantage of using MRI in thermal ablation of hepatic CRC metastases is the ability to help determine the degree of tissue coagulation during energy application by non–invasively measuring temperature in a three–dimensional volume with high spatial and temporal resolution, which allows you to effectively monitor the achievement of tumour coagulation and prevent damage to the surrounding unchanged tissue. To date, several methods of magnetic resonance thermometry have been proposed: T1 relaxation time determination of water molecules, MRI imaging of tissue water molecule diffusion coefficients, proton spin density, proton resonance frequency shift, chemical proton shift, thermosensitive contrast agents, etc. [82–85], but the highest sensitivity to temperature changes in a wide range with an accuracy not differing from that of invasive measurements was demonstrated by the method of determining the proton resonance frequency shift [81]. It should be noted, however, that standard methods of magnetic resonance thermometry, such as the proton resonance frequency shift, are sensitive to radio frequency interference, motion artefacts, and magnetic field drift, but technical improvements have overcome these shortcomings. Even temperature–sensitive sequences have been proposed to adapt the energy release [86] and pulse switches to overcome the interference caused by the use of radio frequencies during MR–RF encoded data collection [87]. Magnetic resonance images after RFA reveal a characteristically altered signal on both T1– and T2–weighted images; the ablation zones also lack gadolinium enhancement. Some researchers pay special attention to the decrease in signal on T2–weighted images as a marker of induced coagulation [88].

Although initial imaging can be a good indicator of ablation adequacy, the resolution and accuracy of modern imaging techniques do not allow for the identification of residual microscopic foci of malignant cells, especially in the periphery of the lesion, where blood circulation is greatest. Their growth will inevitably continue, and this will lead to local recurrence, which dictates the need for long–term monitoring of the effects of RFA using sonography, which is of limited value. The characteristic peritumoral halo is often obscured prior to treatment, and due to the variability of sonographic changes in grey scale, it is impossible to accurately assess induced coagulation [4]. The use of ultrasound contrast can facilitate the differentiation of the avascular ablation zone from tumour recurrence six months after RFA [54].

CT with contrast remains the basis for long–term follow–up in most centres. Coagulated areas without enhancement become more prominent and their edges more distinct two weeks after ablation [89]. Imaging after 6 to 12 months may show marked regression of the lesion and an area of induced coagulation necrosis. Most often, the ablation zone, which is not enhanced by contrast, decreases in volume by about 20%. Often, the coagulation zone is surrounded by a peripheral "halo" rim that is densely enhanced on delayed contrast images; it should not be interpreted as residual tumour, as experimental and clinical studies have shown that this rim is an inflammatory response to thermal cell damage [89]. At the same time, an incompletely cured lesion is most often manifested by a bulky, irregular damage at the edge of the treatment site. When MRI is used for long–term follow–up, attention is paid to the presence or absence of gadolinium enhancement of the ablation zone [89, 90]. Compared to the results obtained within three days after RFA, heterogeneous changes are usually recorded on unenhanced T1– and T2–weighted images. This variability in signal intensity throughout the ablation zone is most likely caused by the uneven evolution of the necrosis area and the organ's response to thermal damage. It should be noted that the excessive variety of images, as well as the many proposed sequences for their registration, virtually eliminates the use of MRI in the long term to prove the effectiveness of treatment. PET/CT with 18–fluorodeoxyglucose is an effective method of long–term follow–up to detect active foci of residual tumour [91]. Treatment is considered adequate if there is no evidence of peripheral tumour recurrence within 12 months.
Thus, the need to ensure effective percutaneous ablation of hepatic metastases of CRC has recently stimulated the development of medical imaging methods mainly in two directions, namely: fusion of images obtained by different methods in real time through the use of specialised software and the development of new ultrasound modalities based on the analysis of certain characteristics of the reflected acoustic signal. The first direction is associated with the best oncological outcomes with very low overall rates of local tumour progression, high technical efficiency, and a significant life expectancy of patients achieved by stereotactic thermal ablation of hepatic metastases, and is in fact a counterweight to resection as a first-line treatment for liver tumours. The second area has been dominated by CE in recent years, which in experimental and clinical studies demonstrates the potential to become the main imaging method for monitoring percutaneous thermal ablation.

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